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The initiation of gait in lower limb amputees: Some related data

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Abstract—The initiation of gait, from balanced standing position to the toe-off of the stance leg, was analyzed in 8 unilateral above-knee (AK) and 10 unilateral below-knee (BK) males amputees. Thirty-one parameters were measured, including ground-foot forces and the movements and timing of hip, knee, and ankle joints. The significant changes from the normal pattern of initiation of gait found in the AK and BK amputees, as well as significant changes between the two amputees groups themselves, are described. The amputees were divided into two subgroups: those who start walking with their prosthesis and those starting with their normal leg. The two groups were compared statistically for each amputation level and all were compared to a normal subjects group. Differences relating to the choice of the swing leg were found. The findings are reported as part of a future databank.

Key words: *above-knee and below-knee amputees, amputation level, gait analysis, prostheses.*

INTRODUCTION

Biomechanical analyses of gait of both above-knee (AK) and below-knee (BK) amputees have been performed by many researchers over the past years. A review by Murdoch (9) described most of the major approaches and techniques still in use and summarized most of the known data. Recent studies have added further information, some aiming at improving prosthetic design and manufacturing (3), others aimed at improving the theoretical analysis (10), and some which attempted to use biomechanics criteria to better fit prostheses (13). Gage and Hicks (5) reviewed the current use of gait analysis in all aspects of prosthetics.

The majority of investigators concentrate their efforts on level walking, which is the steady state, semiautomatic period of walking. Their work led to the development and improvement of a number of highly successful prostheses and auxiliary mechanisms, some of which were described by Murdoch (9). However, biomechanical techniques did not become a standard diagnostic and assessment tool at the clinical and workshop level. There is still an urgent need to identify simple and easily measurable parameters that are significantly affected by various pathologies or misalignments, and the search goes on (15).

So far, the use of gait analysis in the prosthetic workshop and the orthopedic clinic has been limited. One of the major reasons for this situation is the almost exclusive use of level walking data as being representative of gait. It has been shown that similar level-walking kinetics will be achieved with a number of different combinations of active muscles and nerves (11). Many pathologies, prosthetic misalignments, or discomforts might be compensated for, hidden, and/or missed in level-walking gait analysis. The use of electromyography (EMG) at the workshop or clinic, using surface electrodes, is not common and does not offer a real solution, being mostly qualitative and limited in nature (11), while in most cases the use of needle electrodes is not practical outside the hospital and the laboratory.

Investigating gait periods other than level walking might lead to a better understanding and assessment of

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pathologic gait. A number of publications have dealt with special periods or events in gait. A recent publication of this kind (14), analyzes the biomechanics of reactions to impending falls. For safety reasons, this approach cannot be used in the practical situation, but should lead to a better understanding of the locomotor system. Similar approaches can be found in Dickstein *et al.* (4) and Vittas *et al.* (17), both of whom look at standing patterns as an indication for disability.

A handful of researchers have looked into the initiation of gait, the subject of our present work, in the normal, healthy subject (1,2,6,8,12). These studies lead to a better understanding of this important period in gait and create a normal database for future work.

Very few researchers have examined the biomechanics of gait in the amputee performing other than level walking. One such work is by Summers, *et al.* (16), who investigated foot loading during standing. Another is the work published by Vittas, *et al.* (17), who reviewed body sway in BK amputees as a means for assessment of stability.

The purpose of the present study is twofold: first, to start the creation of a database needed for future work concerning the important period of the initiation of gait; and, second, to identify a set of parameters to be used in the clinic and the workshop. We believe that in order for these parameters to be useful in the non-laboratory environment they have to be simple, fast, and inexpensive to measure, as well as easy to interpret, noncontroversial in nature, and causing minimal interference or disturbance to the patient. Following our first paper, in which we described the basic techniques and methods in detail and reported the results from normal male and female subjects (12), the present report introduces the results from AK and BK amputees using an identical approach.

MATERIALS AND METHODS

The initiation of gait (IG) was defined as the time interval between a visual start signal (S) and the toe-off (TO2) of the second foot to leave the floor, defined as the stance leg. Two Kistler force platforms (Kistler Instruments AG) and a Vicon system (Oxford Metrics Ltd., using four TV cameras on-line with a PDP 11/23 computer) recorded the patient's three-dimensional ground-foot forces and lower limb movements, as described in detail by Kirtley, *et al.* (7). A green light served as a start signal. The PDP 11/23 computer was used for data collection and analysis. The patient started the test standing erect, each foot on one force platform. When given a signal, the patient started walking forwards at his own comfortable speed. The swing leg was chosen instinctively by the subject in every test without comments or instructions. All the equipment, methods, and definitions used in this work were identical to those used and described in our previous publication (12).

Ten BK amputees (weighing 788 N on average), eight AK amputees (weighing 738 N on average), and seven normal subjects (weighing 736 N on average), all males, were tested (see **Table 1**). Four tests were recorded for each patient and the last three were evaluated. Thirty-one parameters were measured and analyzed for each run. These parameters, chosen for their relative simplicity and ease of measurement, included the average forces between the start signal (S) and the time of the first measurable reaction (R), the maximal and minimal values of all forces and angular movements in the sagittal plane during the first stride (up to TO2), and the loading and unloading periods of the vertical force between the ground and the feet. Other parameters, such as impulses at various times, moments around the joints, etc., will be the subject of further research. Full details and nomenclature are given in the appendix. The parameters were compared statistically between the various groups, using the nonparametric Wilcoxon rank test.

RESULTS

The results will be presented in two parts: first, the AK amputees, followed by the BK amputees.

Table 1.
Patients and subjects.

Group	Description	n	Age(SD)	Tests
A	normal subjects	7	44(13)	28
AK	above-knee amputees			
B	swing leg = prosthesis	3	—	12
C	swing leg = normal leg	5	—	20
	Total	8	45(15)	32
BK	below-knee amputees			
D	swing leg = prosthesis	5	—	20
E	swing leg = normal leg	5	—	20
	Total	10	53(15)	40

All males, wearing their own, low heel or flat shoes; none using any walking aids.
SD = standard deviation

AK Amputees

A total of 24 tests of AK amputees were analyzed. Three patients systematically used their prosthetic leg as the swing leg to leave the ground first (group B), while the other five used their sound leg as the swing leg (group C).

Twenty-one tests of normal subjects were analyzed for comparison (group A). Results from typical AK amputees, starting with either leg, and a typical normal subject are given in **Figures 1, 2, 3, 4, 5,** and **6**. The detailed results for all the AK amputees, groups B and C, are given in **Table 2**, while a similar list for the normal group, A, appears in **Table 2a** and **Table 2b** of Nissan and Whittle (12).

The parameters which were found significantly different from normal in all AK amputees are described in **Table 3** and **Table 4**. They include smaller peak forces between the stance legs and the ground in all three dimensions, faster unloading of prosthetic stance leg, smaller knee flexion in the stance leg during the whole stride, and diminished ankle flexion. The midswing knee flexion is completely

Table 2a.
Summary of results for above-knee (AK) amputees: forces.

Pt. Code	Amp type leg	Bw N	Swing leg	T1 ms	T2 ms	Fx Max1 %BW	Fx Max2 %BW	Fx Max4 %BW	Fy Av1 %BW	Fy Max1 %BW	Fy Max2 %BW	Fy Max4 %BW	Fz Av1 %BW	Fz Max1 %BW	Fz Max2 %BW	Fz Min %BW	Fz Max3 %BW
IA7	AK-R	750	L	720	1350	6	7	13	5	6	8	10	57	81	99	92	96
IB3	AK-L	900	L	410	1200	−3	9	22	3	4	4	7	24	38	96	92	109
IB4	AK-L	640	R	890	1700	6	5	10	2	5	7	8	58	71	105	94	102
IB8	AK-R	690	R	700	1750	−2	6	22	4	8	3	8	38	56	101	92	98
IC1	AK-R	680	L	830	1560	4	6	11	2	−6	10	11	65	82	100	91	101
IC5	AK-L	660	R	840	1540	8	4	12	4	6	6	5	52	72	102	92	98
IC7	AK-L	700	L	460	1250	0	9	19	4	8	5	9	43	74	93	90	105
IG3	AK-R	880	L	700	1240	12	15	20	3	6	8	9	53	74	102	82	104
Total	AK:	738	—	690	1450	4	8	16	3	5	6	8	49	69	100	91	102
	SD:	93	—	160	200	5	3	5	1	4	2	2	12	14	4	3	4

Directions of forces on the legs:
Fx = positive forward
Fy = positive medially
Fz = positive upwards

Table 2b.
Summary of results for above-knee (AK) amputees: Joint movement.

Pt. Code	Amp type leg	Bw N	Age y	Swing leg	H Max1 deg.	H Max2 deg.	H Max3 deg.	H Max4 deg.	K Max1 deg.	K Max2 deg.	K Max3 deg.	A Max1 deg.	A Max2 deg.	A Max3 deg.	A Max4 deg.
IA7	AK-R	750	41	L	−2	−3	20	6	−2	3	35	1	3	8	−8
IB3	AK-L	900	41	L	−2	6	10	0	0	3	30	0	4	2	−7
IB4	AK-L	640	64	R	4	−3	25	2	0	0	38	—	0	18	−4
IB8	AK-R	690	26	R	−14	—	30	8	0	15	56	3	8	9	−10
IC1	AK-R	680	58	L	−9	−2	30	18	−3	0	38	3	3	16	−8
IC5	AK-L	660	29	R	4	−6	29	30	6	0	50	9	4	8	−8
IC7	AK-L	700	66	L	−10	5	32	12	−1	10	25	0	6	−2	−6
IG3	AK-R	880	35	L	8	—	33	15	−5	0	50	10	2	5	−10
Total	AK:	738	45	—	—	—	26	11	—	4	40	4	4	8	−8
	SD:	93	15	—	—	—	7	9	—	6	10	4	3	6	2

Directions of angular movement:
H and K are positive in flexion; A is positive in dorsiflexion.

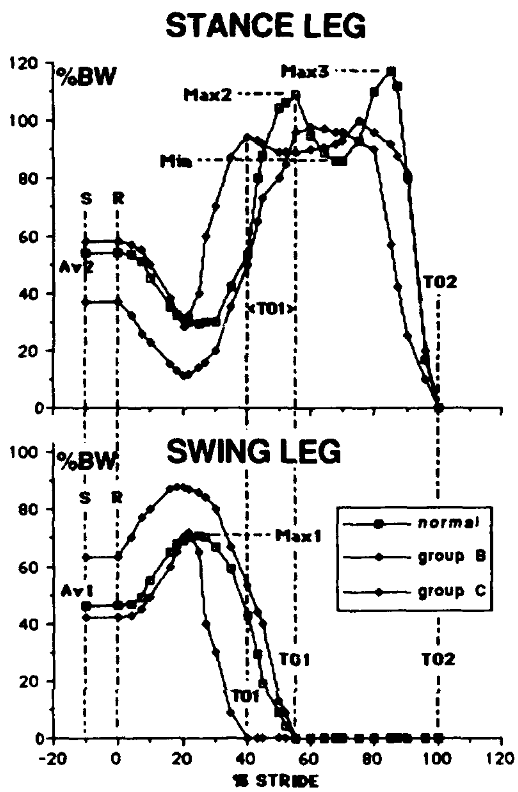

Figure 1.
The vertical force (Fz) in AK amputees and normal subjects.

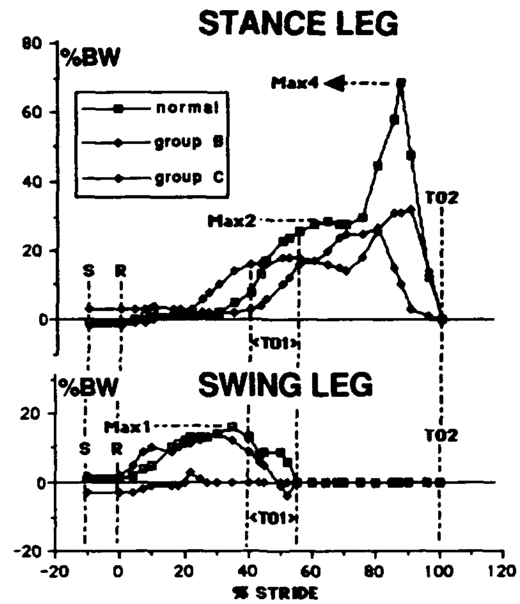

Figure 2.
The fore-aft force (Fx) in AK amputees and normal subjects.

missing in the prosthetic leg. As one would expect, the prosthetic swing ankle is locked.

The statistical significance of the differences between groups A, B, and C are given in **Table 4**. A number of highly significant differences between all AK amputees (groups B and C) and normal subjects (group A) were identified: the fore-aft forces on both legs are much smaller in amputees, compared with normal subjects, and the first peak vertical forces are smaller in the stance legs of amputees in both groups.

The findings related to sub-groups within the AK amputees are of limited value because of the small group sizes. Comparing the AK amputees who used their sound leg as the swing leg (group C) to normal subjects, the only highly significant differences were found in the pre-movement vertical forces (Fz-Av1 and Fz-Av2). The sound swing leg carried much more than half the body weight, thereby reducing the load on the prosthesis. Nonsignificant differences ($p<0.05$) were found in other parameters, such as T, Fx-Max1, Fy-Max2, and K-Max2.

BK Amputees

Thirty tests with 10 BK amputees were analyzed. The amputees' personal data are summarized in **Table 1**. Five patients (15 tests) used the prosthetic leg as their swing leg (group D); the other five used their intact leg as the swing leg (group E). The findings are compared to the normal group A (21 tests) and the two AK amputee groups B and C (24 tests), described previously. Results from typical BK amputees, starting with either leg, and from a typical normal subject are given in **Figures 7, 8, 9, 10, 11**, and **12**. The results from all BK amputees are presented in **Table 5a** and **Table 5b**.

Both BK amputee groups D and E are different from the normal A group in a number of parameters: all ground-foot forces are smaller in the amputee (see **Table 6**). The loading (T1) and unloading (T2) of the amputee's stance leg are shorter than normal. The hips and knees flex to a lesser degree in both legs in BK amputees; the hip hyperextending in the stance legs. The stance leg ankle flexes less towards toe-off. The most significant statistical differences between normal subjects and the BK amputees, the two BK groups themselves, and between BK and AK amputees, are calculated using the Wilcoxon rank-sum test, and are given in **Table 7**.

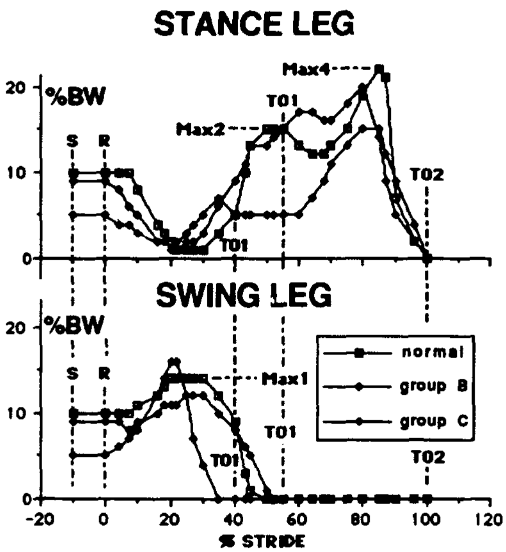

Figure 3.
The mediolateral force (Fy) in AK amputees and normal subjects.

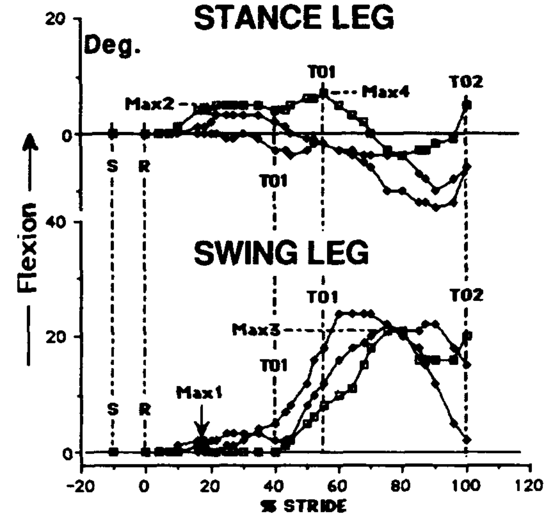

Figure 4.
Flexion-extension of the hips (H) in AK amputees and normal subjects.

DISCUSSION

The results presented in this work seem to be the first concerning the initiation of gait in amputees. The majority of the existing literature and published data deal with standing still or level walking. The assessments of the clinician and prosthetist are based mostly on those two steady-state situations, as described in the introduction. The initiation of gait (IG), occurring whenever there is gait, is too short (total of less than 1 sec) and complex to be analyzed visually, and so far has been ignored. IG is a most interesting period of gait, during which a large portion of the neuromuscular system is necessarily in operation, in contrast with the steady-state level walking, during which only some of the muscles (and the related nerves) have to actively participate (11). It is assumed that pathologies or misalignments which affect level walking in insignificant ways will alter the patterns of forces and movements during the initiation of gait to a much larger extent, thereby enabling a more accurate diagnosis. These changes during IG are expected to occur in the forces, moments, angular and linear movements, and EMG, as well as various combinations of these factors. Some of these parameters are difficult to measure or calculate outside the laboratory or the hospital, while others can be measured using minimal

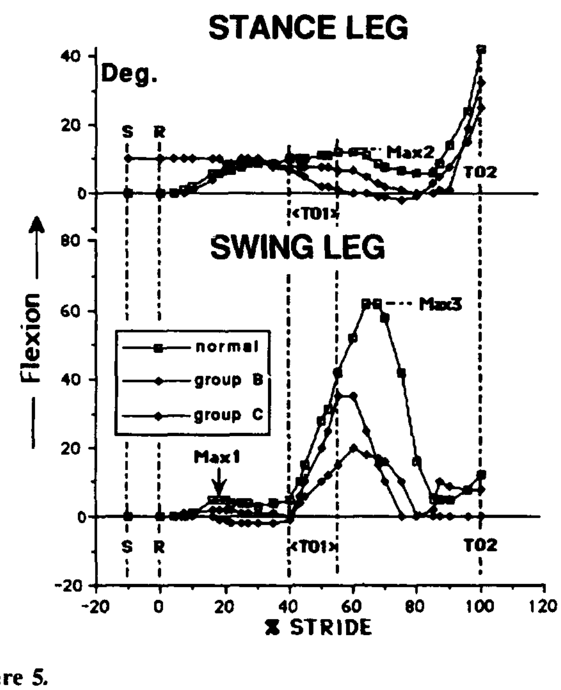

Figure 5.
Flexion-extension of the knees (K) in AK amputees and normal subjects.

equipment in outpatient or workshop conditions by technicians or paramedical personnel. Our research tested the hypothesis that some of the parameters from the second group (i.e., those that are simpler and easier to measure) can be used as indicators of the quality of fitting and the performance of amputees.

The Kistler force platforms can record and analyze data at 500 Hz, but our previous work indicated that a sampling rate of 50 Hz is adequate. The possible spatial errors in the system were described in detail by Whittle (18), who concluded that the combined system, used later in the commercial Vicon system, was "rapid, reliable and easy to use."

AK amputees were found to be significantly different from normal subjects in a number of parameters described above, and in **Table 3** and **Table 4**. Most amputees apparently preferred to unload their prosthesis during the R-S period and to use their sound leg as the swing leg (group C) later on. Slower forward motion resulted from smaller fore-aft forces, causing smaller vertical forces in both legs, especially a smaller peak Fz-Max2, and smaller knee and ankle flexion, compared to normal subjects. The prosthetic stance legs were unloaded significantly faster than in normals. The prosthetic ankle flexed significantly less than the normal ankle in all groups of amputees and in all periods of IG, as expected.

The number of highly significant differences between BK amputees and normal subjects is smaller than the

Table 3.
Summary of major AK amputees' changes from normal, related to gait timing (upper row).

group	leg	param	S	R	TO1		HS1	TO2
		Fx	—	——	smaller	smaller		
		Fy	—	——		smaller		
	STANC	Fz	—	——		trough missing	low peaks	
		H	—	——		——		
		K	—	——	less flexion	——		
		A	—	——		smaller dorsiflexion		
B								
		Fx	—	smaller		——		
		Fy	—	——		——		
	SWING	Fz	—	——		——		
		H	—	——		——	smaller extension	
		K	—	locked		smaller flexion		
		A	—	locked		smaller pf		df
		Fx	—	——	smaller	smaller		
		Fy	—	——		smaller		
	STANC	Fz	overloading	——	smaller	trough missing	low peaks	fast unload
		H	—	locked	extend	——		
		K	—	——	extend	less flex		
		A	—	locked		smaller df		
C								
		Fx	—	——	smaller	——		
		Fy	—	——		——		
	SWING	Fz	underloading	——	smaller	——		
		H	—	——		——		
		K	—	locked		smaller flex		
		A	—	——		smaller pf		

Group B: prosthesis used as swing leg
Group C: prosthesis used as stance leg
pf = plantarflexion; flex = flexion; ext = extension

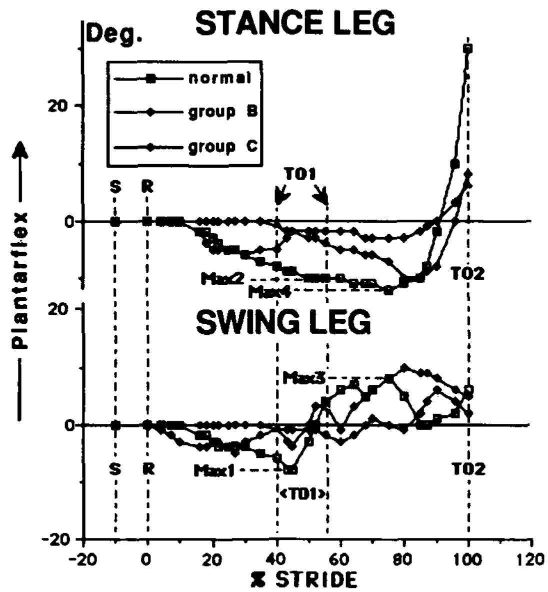

Figure 6.
Dorsiplantar flexion of the ankles (A) in AK amputees and normal subjects.

number found between AK amputees and normals. This result is expected, considering the lesser disability of the BK amputees. The fore-aft forces, the first peak vertical forces at midstance, and the peak plantarflexion in the stance ankles were smaller ($p<0.001$) in the BK amputee, in comparison to the normal subject. The slower pace of unloading the stance leg ($p<0.01$) in amputees, together with the previous results, indicate a tendency for slower and more careful gait. The much smaller BK's second midstance peak vertical force (Fz-Max3) in the sound stance leg (group D, $p<0.001$) is another indication of the same tendency.

Comparison between AK and BK amputees using their prosthesis as the swing leg (groups B and D) result in some differences at the $p<0.05$ level, mainly because of the small sample size. The same results were found between the amputee subgroups C and E, who used their normal legs as the swing leg (see **Table 7**).

Comparing the two BK amputees subgroups in **Table 7**, one finds the normal midstance knee flexion (K-Max2) missing, or even replaced by hyperextension in group E. This easily measurable parameter might be linked directly with major gait deviations and consequently be used by the prosthetist as an indicator for quality of fit. Another difference is the smaller active push-off plantarflexion in the prosthetic stance leg ankles (group E), as can be expected from present day passive prosthetic ankles.

One of the aims of prosthetic fitting is to restore as much of the normal patterns of movement as possible. Our findings, so far, support the hypothesis that some relatively easily measurable parameters in amputees are significantly different from normal subjects during IG. We further assume, and this has yet to be tested, that approaching normal values by changing the prosthetic alignment or components will be followed by improvement in amputee gait.

The correlation between the various parameters measured in the present work (and probably some others to be defined and measured in the future), still needs to be tested. However, we believe that the parameters are interrelated. An improvement in one of the parameters that distinguishes between normals and BK or AK amputees—bringing it nearer the normal value—will indicate a pattern of walking that is nearer the normal and will lead to improvement in the other parameters as well. Further studies are planned, designed to test our last hypothesis. Once proven, this would mean that measurement of very few parameters, possibly even a single force component or angle, will enable an immediate objective assessment of prosthetic fit and alignment. In this case, the equipment

Table 4.
Statistically significant differences between groups.

Groups	Parameter	Signif.
1. N vs AK	Fx-Max1	***
(A vs B+C)	Fx-Max2	***
	Fx-Max4	*
	Fy-Max2	*
	Fy-Max4	*
	Fz-Max2	***
	Fz-Max3	*
	Fz-slope2	**
	K-Max2	*
	A-Max3	*
	A-Max4	***
2. B vs C	Fz-Av1	***
3. A vs B	Fx-Max2	**
(normal leg)	Fy-Max2	**
	Fz-Av1	***
	Fz-Max2	**
	A-Max4	**

N = Group A: normal subjects
AK = above-knee amputees
B = AK, swing leg/prosthesis
C = AK, swing leg/normal leg
* = $p<0.05$; ** = $p<0.01$; *** = $p<0.001$.

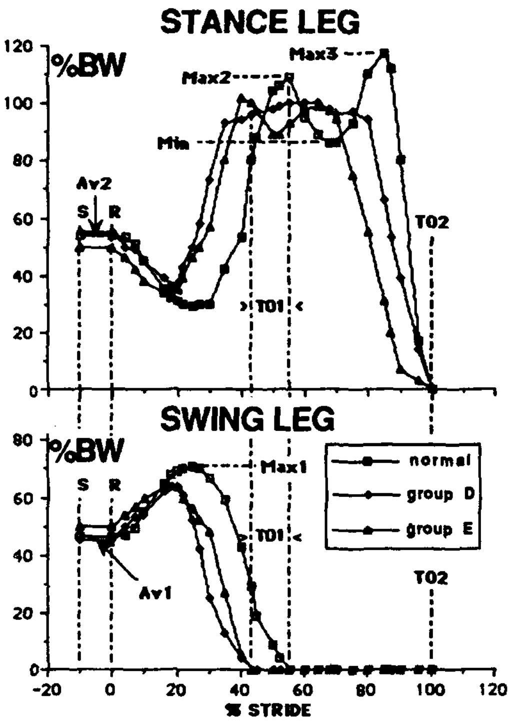

Figure 7.
The vertical force (Fz) in BK amputees and normal subjects.

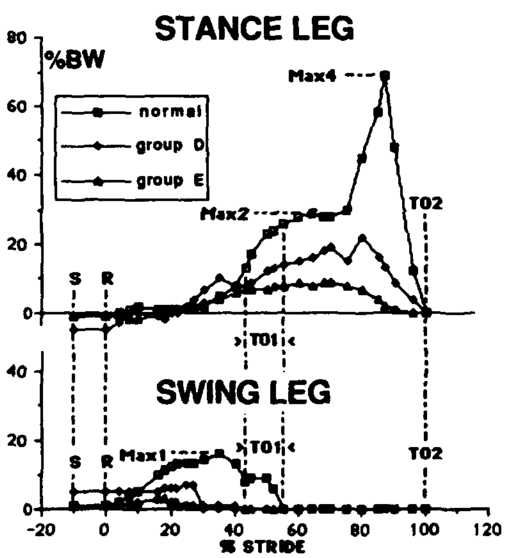

Figure 8.
The fore-aft force (Fx) in BK amputees and normal subjects.

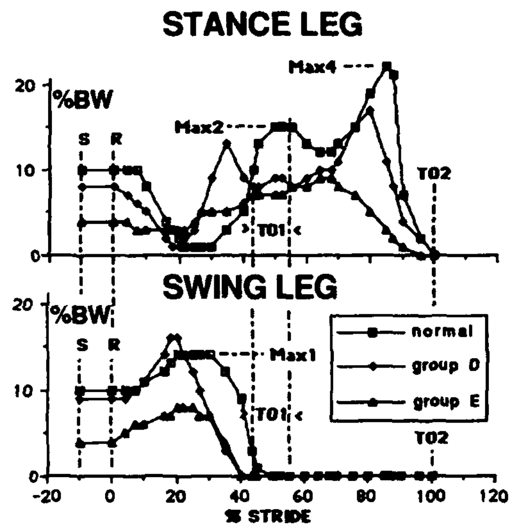

Figure 9.
The mediolateral force (Fy) in BK amputees and normal subjects.

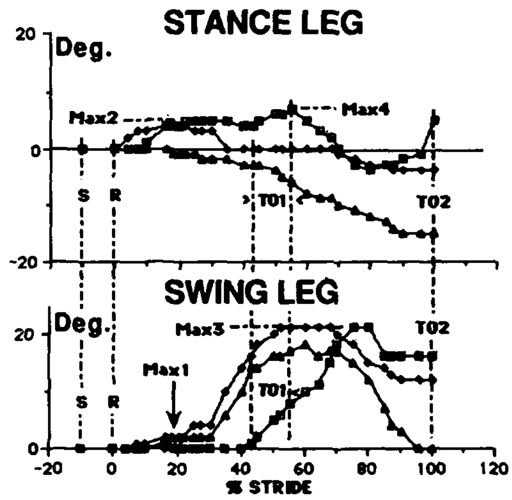

Figure 10.
Flexion-extension of the hips (H) in BK amputees and normal subjects.

needed for prosthetic assessment will be minimal, noninvasive, relatively simple, and inexpensive. We hope this will help in implementing the system in every orthopedic clinic and prosthetic/orthotic workshop.

Considering the spatial resolution and system inaccuracy described by Whittle (18), the major inaccuracy is inflicted by the small sizes of the various groups and the

large number of variables involved. In order to have a useful database, the tested groups have to be enlarged significantly; taking into account the various ankle and knee mechanisms, length of stump, age, alignment, etc. This can be done only by a number of centers cooperating on this subject. The amputees in the present study used a variety of knee, ankle, and foot mechanisms, and the prosthesis alignments varied considerably. This situation limits the scope of the analysis to findings common to all AK or BK amputees, regardless of their specific prosthesis. Testing groups with defined components and/or alignment will enable an assessment regarding the relative merits of the various variables.

Table 5a.
Summary of results for below-knee (BK) amputees: Forces.

						Fx			Fy				Fz				
Pt. Code	Amp type leg	Bw N	Swing leg	T1 ms	T2 ms	Max1 %BW	Max2 %BW	Max4 %BW	Av1 %BW	Max1 %BW	Max2 %BW	Max4 %BW	Av1 %BW	Max1 %BW	Max2 %BW	Min %BW	Max3 %BW
IA8	BK-R	650	L	1000	1700	6	9	10	2	4	8	8	51	66	102	93	101
IB1	BK-L	990	L	740	1530	3	7	10	4	7	9	13	59	83	99	92	96
IB2	BK-R	780	R	600	1330	−3	5	12	4	8	6	9	47	67	98	96	99
IB6	BK-L	810	L	450	1160	−2	9	13	4	6	5	9	34	46	101	91	98
IB7	BK-R	810	L	690	1420	2	7	10	4	6	6	9	48	61	102	90	96
IC6	BK-L	740	L	860	1490	2	6	18	4	6	6	7	45	53	95	86	99
ID2	BK-R	770	L	550	1240	5	12	14	6	10	8	9	39	55	109	89	100
IF3	BK-L	760	L	590	1400	2	9	12	3	4	5	8	37	51	96	91	96
IF7	BK-R	590	L	640	1260	7	10	17	1	7	12	11	56	76	100	90	95
IG2	BK-R	980	L	630	1200	6	13	17	4	7	8	10	47	70	100	90	104
Total	BK:	788	—	675	1390	3	9	13	4	6	7	10	46	63	100	91	98
	SD:	118	—	150	160	3	2	3	1	2	2	2	8	11	4	2	3

Directions of forces on the legs:
Fx = positive forward
Fy = positive medially
Fz = positive upwards

Table 5b.
Summary of results for below-knee (BK) amputees: Joint movement.

					H				K			A			
Pt. Code	Amp type leg	Bw N	Age y	Swing leg	Max1 deg.	Max2 deg.	Max3 deg.	Max4 deg.	Max1 deg.	Max2 deg.	Max3 deg.	Max1 deg.	Max2 deg.	Max3 deg.	Max4 deg.
IA8	BK-R	650	65	L	5	5	18	8	0	0	35	6	0	−8	−3
IB1	BK-L	990	49	L	1	−1	33	4	−1	9	40	−3	7	11	−7
IB2	BK-R	780	65	R	0	4	25	6	0	7	40	6	8	10	−8
IB6	BK-L	810	59	L	0	−3	25	5	1	6	53	2	8	5	−11
IB7	BK-R	810	65	L	3	−2	22	14	1	−7	32	—	0	—	—
IC6	BK-L	740	29	L	−2	5	24	2	−3	7	44	−10	8	12	−11
ID2	BK-R	770	62	L	−2	6	22	6	6	0	43	17	0	6	−1
IF3	BK-L	760	55	L	−1	5	17	3	3	10	30	2	5	11	−10
IF7	BK-R	590	19	L	2	3	28	4	5	−5	37	10	−2	−12	−6
IG2	BK-R	980	57	L	4	−2	30	10	—	−3	35	7	0	10	−7
Total	BK:	788	53	—	2	2	24	6	1	3	39	—	4	5	−7
	SD:	118	15	—	2	3	5	3	3	6	6	—	4	8	3

Directions of angular movement:
H and K are positive in flexion
A is positive in plantarflexion

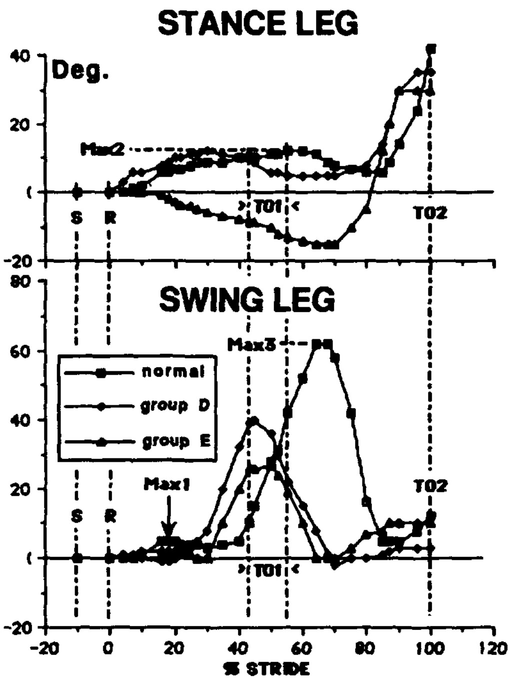

Figure 11.
Flexion-extension of the knees (K) in BK amputees and normal subjects.

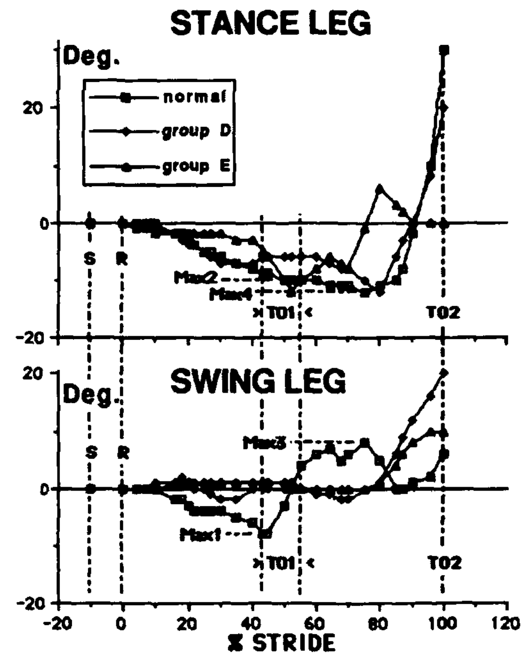

Figure 12.
Dorsiplantar flexion of the ankles (A) in BK amputees and normal subjects.

Further parameters (i.e., the impulses and moments of the various forces) have to be investigated in order to achieve a more profound understanding of the complicated period of the initiation of gait.

There are periods of special interest other than the initiation of gait, such as stopping, climbing stairs, etc. During these periods, different parts of the locomotor system are in extensive use, enabling the identification and analysis of various pathologies and misalignments. These periods have to be looked into in the future. A combined analysis based on level walking, IG, and other periods will enable a more accurate diagnosis and fitting of patients and will improve our theoretical knowledge.

ACKNOWLEDGEMENT

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Table 6.
Summary of major BK amputees' changes from normal, related to gait timing (upper row).

group	leg	param	S	R	TO1		HS1	TO2
D	STANC	Fx	—	smaller		smaller		
		Fy	bigger	—		smaller		
		Fz	—	smaller fast loading	trough missing	low peaks	fast unload	
		H	—	less flex	reduced flex	hyperext		
		K	—	——		extension		
		A	—	——		smaller pf		
	SWING	Fx	—	smaller		———		
		Fy	bigger	——		———		
		Fz	—	——		———		
		H	—	——		smaller flex		
		K	—	locked		smaller flex		
		A	—	——		———		
E	STANC	Fx	—	smaller		smaller		
		Fy	bigger	——		smaller		
		Fz	—	smaller fast loading	trough missing	low peaks	fast unload	
		H	—	less flex		hyperext		
		K	—	less flex	ext	———		
		A	—	——		smaller pf		
	SWING	Fx	—	smaller		———		
		Fy	bigger	——		———		
		Fz	—	——		———		
		H	—	——		smaller flex		
		K	—	less flex		smaller flex		
		A	—	——	variable, depending heavily on make and alignment			

Group D: prosthesis used as swing leg
Group E: prosthesis used as stance leg
pf = plantarflexion; flex = flexion; ext = extension

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Table 7.
Statistically significant differences between groups.

Groups	Parameter	Signif.
1. N vs BK (A vs D+E)	Fx-Max2	***
	Fy-Max4	*
	Fz-Max2	***
	Fz-slope1	*
	Fz-slope2	**
	H-Max2	*
	H-Max3	*
	K-Max1	*
	K-Max2	*
	K-Max3	*
	A-Max4	***
2. AK vs BK	none	
3. D vs E	K-Max2	**
	Fz-Min	*
	Fz-slope2	*
	A-Max4	**
4. B vs D	Fx-Max4	*
5. A vs D (normal leg)	Fx-Max2	***
	Fz-Max2	***
	Fz-Max3	***
	A-Max4	***

N = Group A: normal subjects
AK = above-knee amputees
B = AK, swing leg = prosthesis
BK = below-knee amputees
D = BK, swing leg = prosthesis
E = BK, swing leg = normal leg
* = $p<0.05$; ** = $p<0.01$; *** = $p<0.001$.

APPENDIX

The parameters measured in this test and their notation.

Vertical force
Fz-Av1 = average force up to R, swing leg, in percent BW
Fz-Max1 = maximal force up to TO1, swing leg, in percent BW
Fz-Max2 = maximal force up to TO1, stance leg, in percent BW
Fz-Min = minimal value in the mid-stance trough, in percent BW
Fz-Max3 = maximal force before TO2, stance leg, in percent BW
Fz-Slope1 = maximal loading slope in stance leg, in N/sec
Fz-Slope2 = maximal unloading slope towards TO2, in N/sec

Fore-aft force
Fx-Av1 = average force up to R, swing leg, in percent BW
Fx-Max1 = maximal force up to TO1, swing leg, in percent BW
Fx-Av2 = average force up to R, stance leg, in percent BW
Fx-Max2 = maximal force up to TO1, stance leg, in percent BW
Fx-Max4 = maximal force towards TO2, in percent BW

Mediolateral force
Fy-Av1 = average force up to R, swing leg, in percent BW
Fy-Max1 = maximal force up to TO1, swing leg, in percent BW
Fy-Max2 = maximal force up to TO1, stance leg, in percent BW
Fy-Max4 = maximal force towards TO2, in percent BW

Hip, Sagittal plane, in degrees. Positive values are for extension
H-Max1 = maximal change up to TO1, swing leg
H-Max2 = maximal change up to TO1, stance leg
H-Max3 = maximal change towards TO2, swing leg
H-Max4 = maximal change in mid stance, stance leg

Knee, Sagittal plane, in degrees. Positive values are for flexion
K-Max1 = maximal change up to swing-phase flexion, swing leg
K-Max2 = maximal change up to TO1, stance leg
K-Max3 = maximal swing phase flexion

Ankle, Sagittal plane, in degrees. Positive values are for plantarflexion
A-Max1 = maximal change up to TO1, swing leg
A-Max2 = maximal change up to TO1, stance leg
A-Max3 = maximal change in swing phase
A-Max4 = maximal change in stance phase

Cycle time, in miliseconds
T1 = from R to TO1
T2 = from R to TO2

The maximal values in all cases are the maximal absolute values.

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Effect on gait using various prosthetic ankle-foot devices

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Abstract—Five different commonly prescribed ankle-foot devices for below-knee prostheses were tested for effects on gait: SACH, SAFE, SEATTLE, SINGLE AXIS, and MULTIPLE AXIS. Subjective ratings by the amputees served to resolve which physical variables determined the preferred ankle-foot device. Ratings were related to age, body weight, length of residual limb, and ratio of stride frequency to stride length. Distinctions in performances were derived from analyses of anterior-posterior angular accelerations occurring in the prosthesis immediately following heel strike. The accelerations were characterized as a damped oscillatory waveform. These objective findings, when related to the subjective ratings, showed that the amputees preferred devices which developed the lesser shock and greater damping.

Key words: *ankle-foot device, below-knee prostheses, gait analysis, stride frequency and length, unilateral amputees.*

INTRODUCTION

Early development efforts in lower limb prostheses placed major emphasis in the area of socket design and limb-socket relations and only a minor emphasis on distal components. Subsequent to progress in the socket design, the focus changed biomechanical interest to the distal portions of the prosthesis and to ambulatory performance with existing ankle-foot devices. As a result, there has been a proliferation of designs attempting to meet various requirements. Requirements vary because of differing physical characteristics and activity objectives of patients.

Ground reaction forces are transmitted through the ankle-foot assembly to the prosthesis, the residual limb, and to the rest of the body. How and where these forces act are often conditioned by the characteristics of the ankle-foot assembly. The issue is compounded by the varying nature of the substratum which in turn affects the ambulatory function and comfort of the patient.

While substantial progress has been made in the design of ankle-foot devices, sometimes indications and contraindications for their use have not been clear. This has left clinicians without a sound base or reason for a choice. The purpose of this study was to analyze the effect on gait of five commonly prescribed devices and to suggest additional guidelines for their selection and prescription.

METHOD

Test subjects

The test sample included 19 unilateral amputees who were active community ambulators, had no balance problems, and were compliant with test procedures. The 15 males and 4 female below-knee amputees averaged 46.3 years of age ranging from 24 to 72. Average height was 177.7 cm ranging from 160 to 191 cm. The average weight was 86.8 Kg ranging from 57 to 116. Residual limb lengths averaged 14.6 cm ranging from 8 to 23 cm; in percent of height, these lengths averaged 8.2 and ranged from

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4.5 to 12.8 percent (see **Table 1**). Eight subjects were left amputees and 11 were right. One subject's test data were not included in the sample. This subject was a diminutive female, 166 cm tall, 60 Kg in weight, and 52 years of age. She was unable to perform tests with the multiple axis device because she said "it was too heavy and cumbersome."

Ankle-foot devices

The five ankle-foot devices used in this study were: SACH, SAFE, SEATTLE, SINGLE AXIS (Otto Bock 1H31), and MULTIPLE AXIS (Greissinger), and were presented in the order listed. The order had to do with the length of pylon, with the SACH requiring the longest and the MULTIPLE AXIS requiring the shortest. The devices were selected, adjusted, and aligned in accordance with the manufacturers' specifications. Taken into consideration were the subjects' weights, ages, and nominal activity levels.

Each subject was fitted with a patellar-tendon bearing (PTB) socket constructed according to standard procedures. Construction was of a plastic laminate with a Pelite insert and suprapatellar cuff. The Otto Bock or United States Manufacturing Company adjustable pylon was used to allow easy exchange of ankle-foot devices. Each prosthesis was dynamically aligned with each of five different ankle-foot devices prior to testing. Lightweight, three-quarter inch heel, crepe sole (Hush Puppy) footwear were applied.

Procedure

The subjects were attired in comfortable foot wear, shorts, and shirt. Body-mounted instrumentation included bilateral knee electrogoniometers, bilateral foot switches which covered the heel and sole areas, a triaxial accelerometer mounted on a snug-fitting hip-waist band, and a lightweight backpack containing an electrical junction box.

Wires from the body-mounted instrumentation were plugged into the back pack. Signals were conducted to the signal conditioner through a festooned flat multiconductor cable supported by an overhead carrier. The force needed to pull the cable was about one newton. Signals were conditioned by amplification, filtering with a 33 Hz roll off, and conversion from analog-to-digital. Sixteen channels of data were sample at 60 times per second and stored in the microcomputer and transferred to diskettes for storage and processing.

A lightweight accelerometer pack was secured firmly

Table 1.
Physical characteristics of test subjects.

Subj	Ht cm	Wt Kg	Age yrs	Sex	LRL cm	Pers A-F D
1	191	88	30	M	17	SAFE
2	185	91	72	M	10	SACH
3	160	57	64	F	13	SAFE
4	178	116	52	M	19	SAFE
5	187	107	24	M	17	SAFE
6	178	80	40	M	8	SEATTLE
7	183	111	59	M	19	SACH
8	165	74	70	M	12	M.AXIS
9	180	77	33	M	23	SAFE
10	183	83	53	M	13	SACH
11	188	102	37	M	15	SEATTLE
12	180	102	43	M	17	M.AXIS
13	173	87	39	M	11	SEATTLE
14	160	84	60	F	14	SACH
15	165	66	51	F	11	SACH
16	188	93	33	M	19	SACH
17	180	85	38	M	16	SEATTLE
18	165	68	42	F	14	SAFE
19	188	79	39	M	9	SACH

LRL = Length, residual limb
Pers A-F D = type of personal ankle-foot device

to the lateral side of the socket. This assembly contained five miniature accelerometers (Entran EGAXT): one pair sensed anterior-posterior acceleration, another pair sensed medial-lateral acceleration, and the fifth sensed acceleration parallel to the longitudinal axis of the prosthesis. The paired accelerometers, spaced 10 cm apart, provided differential signals to register angular accelerations calibrated in radians per second squared.

Forward velocity of the body was measured with a tachometer consisting of a low inertia DC generator mounted on a stationary base at one end of the walkway. A braided steel cable 0.014 inch in diameter connected the tachometer to the waist band on the subject. A weight and pulley arrangement maintained constant tension of about one newton on the cable.

Whole body accelerations were monitored with a miniature triaxial device used in a biaxial mode to register longitudinal (forward) and vertical axes of motion. The device was mounted in a two-axis gimbal arrangement secured to the sacro-lumbar area on the hip-waist band. The gimbal was stabilized to the direction of forward progression by the tachometer cable.

The walkway was a level, vinyl tiled floor 7.5 m long and 2 m wide. In addition, two specially constructed, moveable sections, 3 m long and 75 cm wide, were used to provide nonlevel walking surfaces. Laid on the walkway, they provided a surface tilted 7 degrees from horizontal laterally. In one mode, the subjects walked with the prosthesis on the "uphill" side (inclined) and in the other mode they walked with the prosthesis on the "downhill" side (declined). When the same sections were placed on supports, the assembly provided an up ramp and down ramp surface tilted 7 degrees from horizontal.

Test protocol

Each subject was asked to wear the test prosthesis at least 2 hours per day during the week prior to test appointments to become accustomed to the assembly.

Seven test walks were recorded with each of the five ankle-foot devices: 1) three tests on the level surface consisted of "usual," "fast," and "slow" speeds. These speeds were arbitrary and subjectively chosen by each subject; 2) one test each at "usual" speed with the surface inclined laterally and declined laterally; and, 3) two tests at "usual" speed on the ramp assembly to yield two ascents and two descents. In subsequent data processing the two ascending and two descending test data were averaged to yield one representative up ramp and one down ramp performance.

Subjects completed a one page questionnaire: 1) rate the ankle-foot devices according to poor, fair, good, excellent, and superior; and, 2) write a brief comment about each device.

Data treatment

Data collected from each test condition were analyzed by gait cycles. The beginning of the prosthetic limb support period was defined as the beginning of the cycle. Primary indication of this event was the signal from the accelerometer mounted on the prosthesis which sensed acceleration parallel to the longitudinal axis of the prosthesis. Confirmation of this event was closure of the heel switch.

Depending on the speed of walking, three to five gait cycles were obtained for analysis for each test condition. Gait cycles not included were those clearly accelerative at initiation of gait and decelerative at cessation of gait.

Two of the gait variables measured were the forward velocity, V, and the duration of the gait cycle, t. Stride frequency, F, was calculated from the reciprocal of the cycle time: $F = 1/t$. Stride length, L, was calculated as the product of the average velocity, Va, and cycle time, T: $L = Va \times t$. The ratio of stride frequency to stride length was calculated by dividing stride frequency, F, by stride length, L.

Body movement waveforms were quantitated using a Fourier series technique. Fourier coefficients for 18 harmonics were calculated for the anterior-posterior (AP) and medial-lateral (ML) angular accelerations of the prosthesis, and coefficients for 12 harmonics were calculated for the fore-aft and vertical accelerations of the body and for the tachometer. Coefficients for each waveform were averaged from the selected gait cycles to represent a given test condition. Waveforms for the AP and ML angular accelerations of the prosthesis were subsequently synthesized from the Fourier coefficients.

The angular acceleration of the prosthesis at the beginning of the support phase was characterized mathematically as a damped oscillatory wave. Regression analysis was used to determine the best estimate of initial acceleration and the damping factor.

To minimize variance due to differences in height, weight, and length of residual limb, certain normalizing procedures were introduced:

a. To compare results among subjects of differing weight, relative weights of subjects were defined as the actual weight divided by ideal weight, Wa/Wi. The commonly accepted formulas (metricated) for ideal weight

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are the following:

Males: Wi = 51 + (H − 152)

Females: Wi = 48 + 0.83(H − 152)

where:

Wi = ideal weight in Kg

H = height in cm

b. Normalized residual limb length was determined as a percentage of body height and termed RL.

c. To normalize leg-to-substrate angulations as they relate to stride length for subjects of differing stature, stride length, L, was divided by body height, H, to yield the L/H ratio.

Harmonic ratios were calculated in the manner described by Robinson (5) using the Fourier coefficients for whole body movements, namely, the longitudinal and vertical accelerations and forward velocity. In addition, the external work and efficiency of locomotion was calculated in the manner described by Cavagna (3) using the body accelerations and forward velocity, for the three speeds of level walking and for the lateral incline and decline test conditions. Work and efficiency were not calculated for the ramp tests because existing programming was designed for level walking. To modify programming to accommodate reliably the ascent and descent of ramps was deemed not warranted.

RESULTS

Subjective rating

The subjects rated each ankle-foot device in terms of excellent, superior, good, fair, and poor. The results were as follows: 1) SACH: 18 of 19 good to poor equally divided, none superior; 2) SAFE: 14 of 19 good to excellent, two superior, and two poor; 3) SEATTLE: 9 of 19 excellent, three superior, six good and fair, and one poor; 4) SINGLE AXIS: 14 of 19 good and fair, four poor, and one superior; and, 5) MULTIPLE AXIS: 16 of 19 excellent to fair, one superior and one poor. See **Table 2** for tabulated results.

Table 2.
Summary of subjective ratings of ankle-foot devices.

	SACH	SAFE	SEATTLE	S.AXIS	M.AXIS
Superior	0	2	3	1	1
Excellent	1	6	9	0	5
Good	6	8	4	7	7
Fair	6	1	2	7	4
Poor	6	2	1	4	1

Written comments provided reasons why individuals like or disliked given devices. From the varied responses, the following generalities were drawn:

a. SAFE and SEATTLE: These drew most of the favorable comments such as flexible, springy, comfortable, energy-saving, good on slopes and inclines. Older subjects, however, complained of balance and control problems and stated that they were being thrust forward.

b. SACH: The common complaint with this device was that it was too stiff. Heavier subjects objected less.

c. SINGLE AXIS and MULTIPLE AXIS: These devices drew varied reactions. Many opined that the devices were good for multi-terrain applications, with the MULTIPLE AXIS being favored over the SINGLE AXIS. The SINGLE AXIS device was said to lack "side to side" action, and some complained of not feeling secure. Shorter, slimmer individuals complained that the MULTIPLE AXIS was cumbersome. Many objected to the abrupt stop in movement during single support. This was expressed variously as two-point roll over, a clopping sensation, a sudden stop, or a knee-jerking effect. Many who liked these devices reported difficulty in getting accustomed to walking with them. Reactions regarding the resistance varied from too soft to too stiff.

Subjective rankings

Trends noted suggested that the subjective ranking order of the ankle-foot devices was influenced by four independent variables: age, relative weight, length of residual limb, and ratio of stride frequency to stride length. By applying discriminating thresholds to ankle-foot devices, a good match with subjective rankings was achieved:

1. Age, in years: if less than 39, then SEATTLE; if between 39.5 and 52, the SAFE; if over 52.5, the SACH.
2. Wa/Wi, actual weight divided by ideal weight: if less than 1.11, then SEATTLE; if between 1.12 and 1.21, the SAFE; if more than 1.22, then SACH.
3. RL/HT, residual limb length as percent height: if less than 7.2, then MULTIPLE AXIS; if between 7.3 and 9.5, the SEATTLE; if over 9.6, the SAFE.
4. F/L, stride frequency divided by stride length: if less than 0.6, then SAFE; if between 0.61 and 0.75, the SEATTLE; if more than 0.76, then SACH.

From these, the following may be inferred: 1) SEATTLE, if young, lightweight, medium length residual limb, and average step length; 2) SAFE, if middle-aged, slightly overweight, long residual limb, and tendency to long step length; and, 3) SACH, if old, overweight, medium to long residual limb, and tendency to short step length.

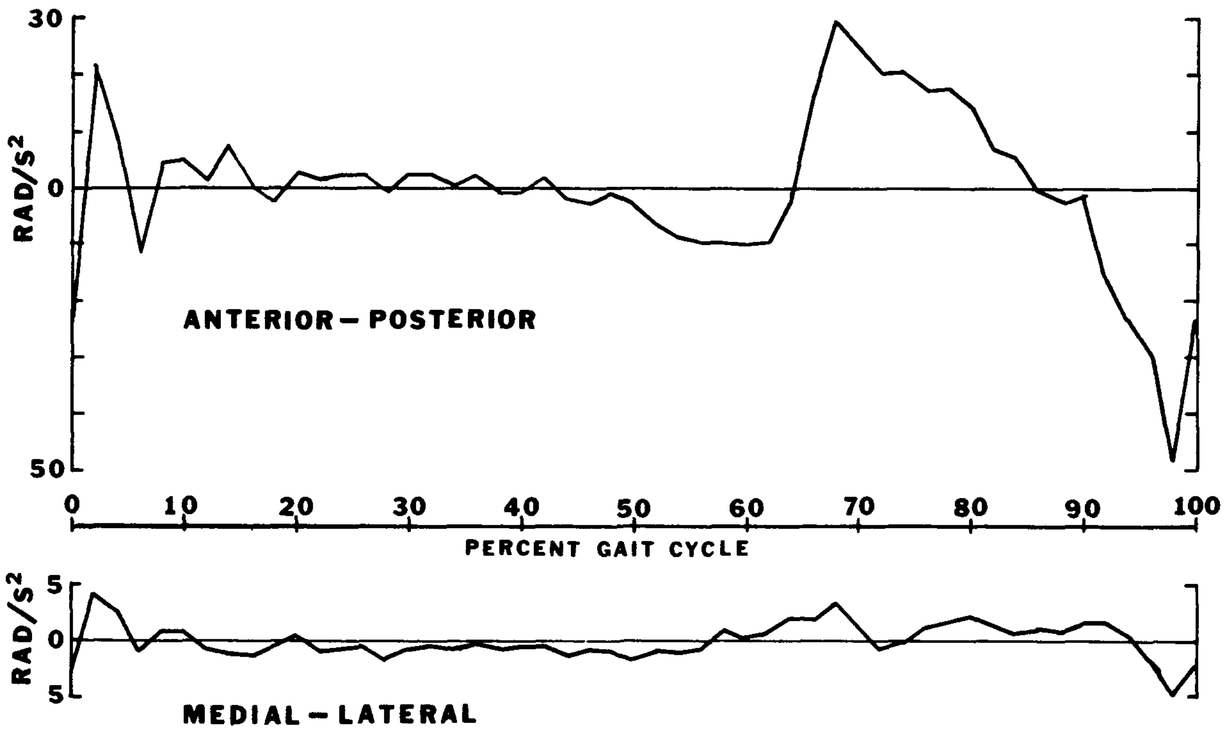

Figure 1.
Angular acceleration of prosthesis as a function of gait cycle.

Prosthesis angular acceleration

Angular acceleration curves were synthesized at two percent gait cycle intervals. Representative waveforms for the anterior-posterior and medial-lateral directions may be seen in **Figure 1**.

a. For all feet at initial stance, the curves typically showed an oscillatory waveform which subsided within two to three cycles (within 20 percent of the gait cycle). During this period, the magnitude of the ML acceleration ranged from about one-tenth to about one-fifth the magnitude of the AP acceleration.

b. At mid-stance (from 20 to about 40 percent gait cycle), the waveforms wavered a small amount.

c. At terminal stance (about 40 to 70 percent gait cycle), the waveforms departed in direction and between 60 and 70 percent gait cycle quickly reversed direction.

d. Between 70 and 100 percent gait cycle, the curves included a substantial acceleration and deceleration to characterize the swing phase.

The amount of "roughness" in the waveforms, from 20 to 40 percent gait cycle, did not relate to quality of performance. The area under the curve from 40 to about 65 percent gait cycle related to velocity of walking but did not discriminate among ankle-foot devices. The curves during swing phase were not studied intensely because deviations in this segment of the waveforms were attributed to characteristics of the suspension system.

AP angular acceleration

The initial stance portion of the AP waveform was used to discriminate performances among the ankle-foot devices. The waveform damped logarithmically:

$$A = Be^{-dp}$$

where

A = Angular acceleration in r/s^2
B = acceleration at zero percent gait cycle
d = damping factor
p = percent gait cycle

This relation was used to define "shock." By assuming a point 8 percent into the gait cycle and applying the initial acceleration and rate damping, values were calculated to represent shock for each performance.

Shock was related to F/L (stride frequency to stride length ratio) and L/H (stride length divided by height). The relation is demonstrated as a nomograph in **Figure 2** to offer a comparison of the five ankle-foot devices. The

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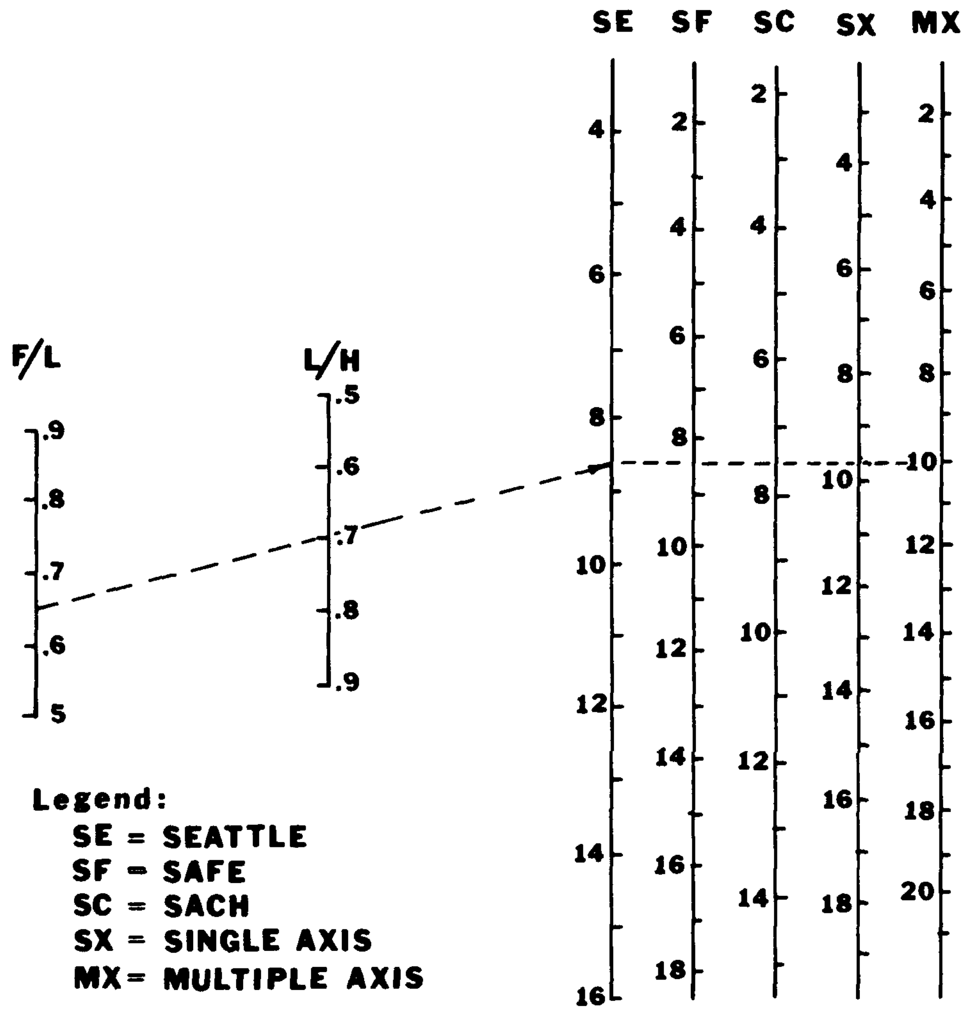

Figure 2.
Nomograph for comparing shock among ankle-foot devices.

SEATTLE, SAFE, and SACH showed the lowest values, while the SINGLE AXIS and the MULTIPLE AXIS devices showed the highest. The dashed line illustrates use of the chart. By example, with F/L = 0.65 and L/H = 0.7, the resulting shock values are the following: SEATTLE = 8.6, SAFE = 8.3, SACH = 7.5, SINGLE AXIS = 9.7, and MULTIPLE AXIS = 10.0. The accuracy is about ± 1 rad/s^2. Comparison of magnitude of shock with subjective ratings showed that the device exhibiting the lease shock corresponded to the subjective preference by the amputee.

Damping

The SACH, SAFE, and SEATTLE devices provided the highest damping for most subjects. Relative weight and relative length of residual limb determined which provided the highest damping for a given individual. Distinction between the three was possible by the following expression:

$$T = Wa/Wi + 0.095(RL)$$

where T is a threshold value. The boundary between SACH and SEATTLE was 1.9, meaning that if the value, T, was less than 1.9, then SACH provided the higher damping. The boundary between SEATTLE and SAFE was 2.2, meaning that if the value, T, was more than 2.2, then SAFE provided the higher damping (see **Figure 3**). The above correctly classified 17 of the 19 subjects.

Oscillatory frequency

The frequency of the shock oscillation averaged about 9 Hz and was governed more by characteristics of the individual than by the ankle-foot device. Among individuals, average frequency ranged from 6.5 to 11.0 Hz. The frequency fs related to relative length of residual limb, RL expressed in percent height, and height H in cm, as follows:

$$fs = 7.1 + 0.23(RL) + 0.054(178 - H)$$

This correctly expressed the frequency within 1 Hz in 14 of 19 subjects. The largest error was 1.6 Hz among the remaining five subjects.

General performance

The AP damped oscillatory angular acceleration or shock of the prosthesis after heel strike related linearly to the velocity of walking. The slopes of the lines of regression were all positive and differed for each device. Briefly summarized:

a. Sensitivity to velocity, in r/s^2 per m/s: SACH = 5.47, SEATTLE = 5.90, SAFE = 8.35, SINGLE AXIS = 10.15, and MULTIPLE AXIS = 11.52.

b. Shock, in r/s^2 translated to velocity of 1.0 m/s: SACH = 7.9, SEATTLE = 8.8, SAFE = 9.0, SINGLE AXIS = 10.3, and MULTIPLE AXIS = 10.8.

c. Highest correlations were on the level surface and the lowest on the lateral incline.

d. Based on significance of correlation, SINGLE AXIS showed the least variability and SEATTLE the most. Results of the correlations are as follows:

Level

Values averaged from the three test speeds characterized performances on the level surface. All correlations were significant. The SEATTLE had the flattest slope and the MULTIPLE AXIS the steepest. The SACH, SAFE, and SEATTLE developed the lesser AP shock while the SINGLE and MULTIPLE AXIS devices developed the greater. (**Figure 4a**).

Lateral incline

The lateral incline test condition was with the prosthesis on the "up hill" side. Results were more diverse than on the level surface. The SINGLE and MULTIPLE AXIS devices showed low correlations while SACH, SAFE, and SEATTLE were not significant. The SACH showed the flattest slope and the MULTIPLE AXIS the steepest. (**Figure 4b**).

Lateral decline

The lateral decline test condition was with the prosthesis on the "down hill" side. SACH and SINGLE AXIS showed good correlations while SAFE, SEATTLE, and MULTIPLE AXIS were not significant. (**Figure 4c**).

Up ramp

Slopes for SACH, SAFE, and SEATTLE were approximately the same. The SACH, SEATTLE, and MULTIPLE AXIS showed good correlations while SAFE and SINGLE AXIS were not significant. The slope for MULTIPLE AXIS was the steepest. (**Figure 4d**).

Down ramp

The down ramp performance were more diverse than up ramp. SAFE, SINGLE AXIS, and MULTIPLE AXIS

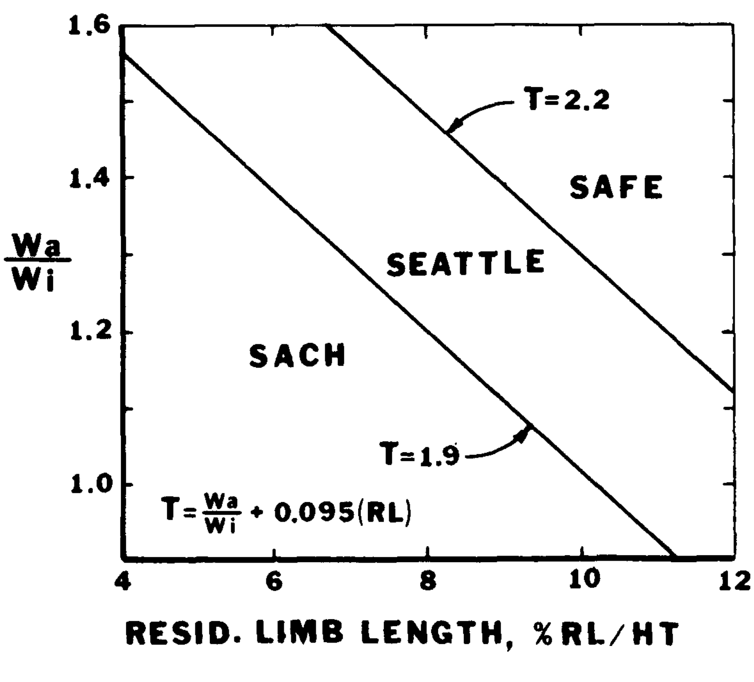

Figure 3.
Boundaries delineating SACH, SEATTLE, and SAFE for highest damping.

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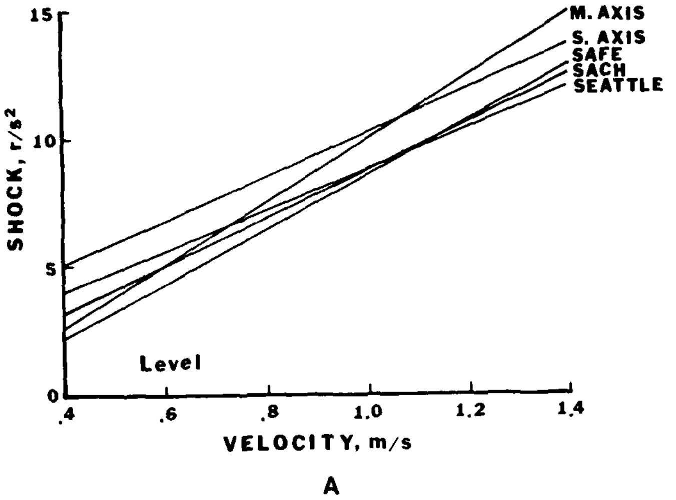

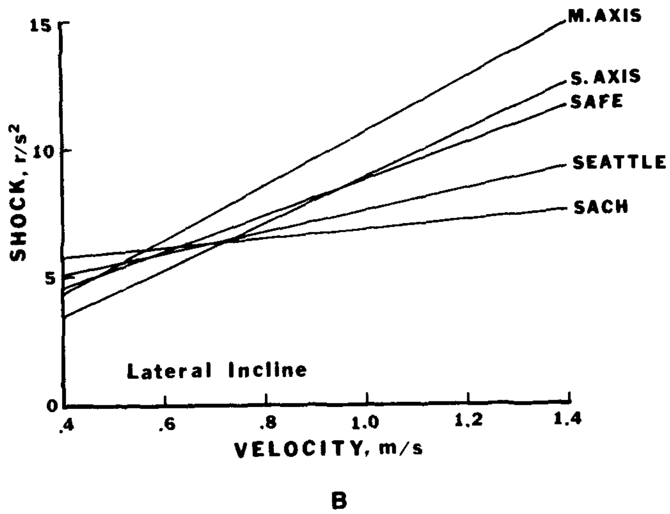

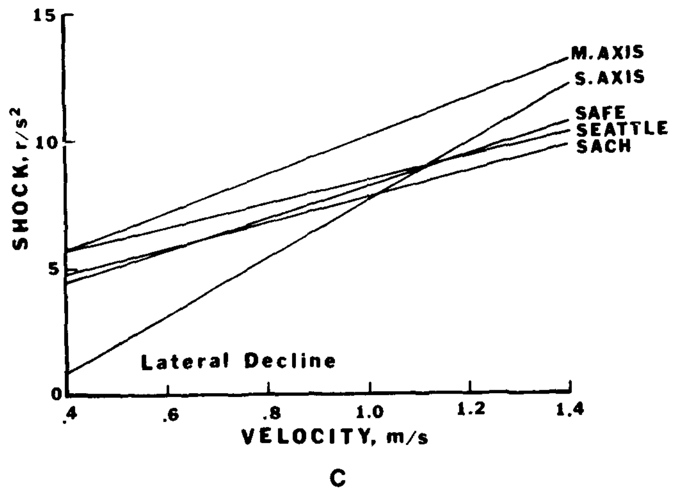

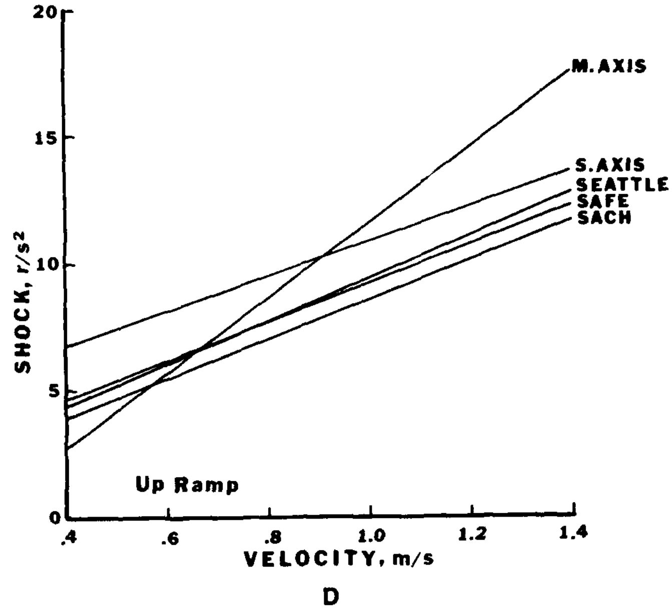

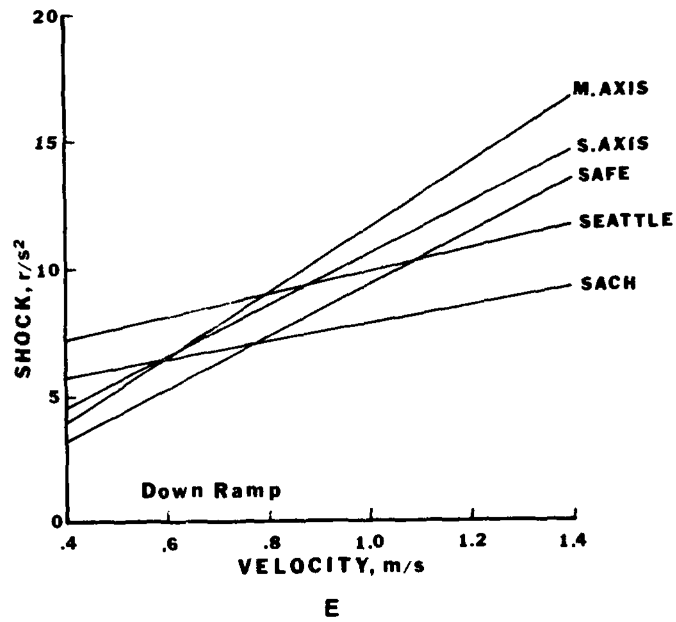

Figure 4.
Regression lines of shock versus velocity for ankle-foot devices. a) level, b) lateral incline, c) lateral decline, d) up ramp, and e) down ramp.

showed good correlations while SACH and SEATTLE were not significant. The slopes for SACH and SEATTLE were the flattest and for MULTIPLE AXIS the steepest. (**Figure 4e**).

Velocity versus age

Average velocity Va in m/s of walking experienced under different test conditions varied inversely with age A, in years. The relation was the following:

$$Va = 1.36 - 0.009A$$

The relation may be seen in **Figure 5.**

F/L and L/H interdependence

The magnitude of shock related to two derived variables, F/L (stride frequency to stride length ratio) and L/H (stride length to height ratio). These variables offered insight into the effect of age on locomotion control strategies involving trade-offs between stride length and stride frequency, when accommodating changes in speed and in walking on unlevel surfaces. Performance characteristics of the nominally younger and older subjects tended to differ in a number of respects.

Younger. To manage the slow speed of walking on a level surface, stride length usually increased and stride frequency decreased. In accommodating the incline-decline surfaces, a comfortable F/L ratio was maintained to achieve a "usual" speed. In walking up the ramp, the comfortable F/L ratio was maintained, but the stride length was decreased. In walking down the ramp, the F/L ratio was increased and the stride length (L/H) was decreased.

Older. The older subjects' performances showed less ordered relations between stride frequency and stride length. Many test subjects over 50 years of age demonstrated an increased F/L ratio at the "slow" speed. In accommodating the unlevel surfaces, L/H was reduced and the F/L increased drastically. Moreover, considerable scatter in F/L and L/H relations occurred in the four tests involved with walking on unlevel surfaces.

Miscellaneous

Harmonic ratios, which indicate degree of bilateral asymmetry, were in the order of 1.7 and ranged in isolated cases from about 0.9 to 4.4. By comparison, the harmonic ratios of healthy individuals are in excess of 3.0. Harmonic ratios did not correlate with angular accelerations of the prosthesis. Moreover, no significant relations were found between the harmonic ratios and subjective ratings by the amputees.

External work of locomotion varied both among and within subjects. Among subjects, the work ranged from about 0.35 to 0.7 J/Kg−m walked. Within subjects, work ranged about ±20 percent attributable to the effects of

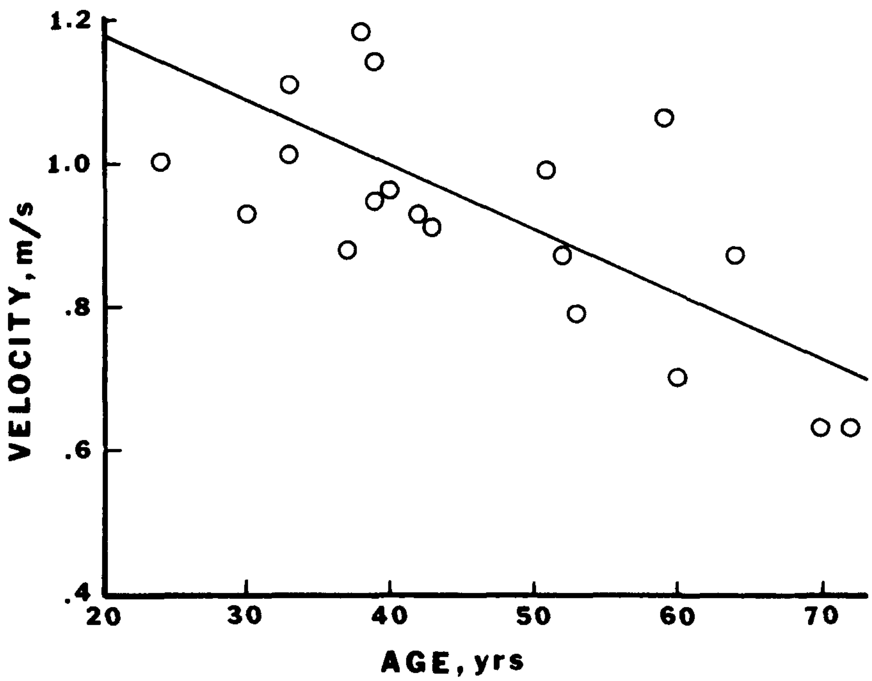

Figure 5.
Relation of walking speed to age.

ankle-foot devices. Correlation between subjective ratings of the different prosthetic feet and numerical values for work was not significant.

DISCUSSION

Not surprisingly, subjective responses varied considerably. This could be expected because of variation among the subjects such as age, physical characteristics, experience with prostheses, influence of personal prosthesis, and differing abilities to verbalize distinctions among the devices tested. However, these responses were invaluable in the effort to determine which measured variables served to discriminate performances in a manner consistent with opinions expressed by the amputees.

The subjective rankings of the ankle-foot devices led to distinctions among the devices by virtue of age, weight, length of residual limb, and ratio of stride frequency to stride length. These results may be due partly to design by the developers of certain components (1,2) and partly by age, physical condition, and attitude of the amputee. Regarding the older subjects' reactions to the SAFE and SEATTLE devices, one might conclude that older people don't want to be hurried! While it is beyond the scope of this report to try to delineate cause and effect for these observations, the observations should serve as additional guidelines in the prescription of ankle-foot devices.

Inferred from the amputees subjective responses was that the mechanism producing the lease shock represented the preferred device. One exception was the SACH. The SACH demonstrated lower shock and higher damping values than others in many cases, yet many subjects rated the SACH as inferior. Written commentaries suggested that the SACH was not as well suited for activities such as running or dancing. While vigorous activities were not on the test protocol of this project, the subjective observation should serve as a prescription guideline.

Presumably the amputees perceived shock as merely a perturbative impact even though shock consisted of three elements: amplitude, frequency, and rate of damping. The effect of shock on the residual limb, as defined in this report, is the product of torque and time, known in mechanics as impulse. Neither the magnitude of the initial angular acceleration nor the rate of damping are adequate by themselves to characterize impact. Both intensity and duration are required. The choice was made to calculate the magnitude of the oscillatory acceleration at 8 percent into the gait cycle. The choice of 8 percent was purely arbitrary. Proceeding beyond the initial foot strike provided the means to register the effects of the magnitude of the initial acceleration, the degree of damping, and a time relation to quantitate the perturbation.

Damping of shock is not a simple issue. Damping occurs during the period when the stump seats into the socket and is partly due to the characteristics of ankle-foot device and partly due to the progressively increasing coupling of the prosthesis to the stump. The reason that comparisons between devices could be made was because the socket and suspension remained the same for each individual for each of the five ankle-foot devices. Hence, differences could be attributed to the effect of a given ankle-foot device on each individual.

A trend noted was that shock frequency increased with length of residual limb. This is may be partly attributable to the longer limb developing a longer lever arm (purchase) and thereby increasing the "stiffness" of the system.

The ratio F/L (stride frequency to stride length ratio) was sensitive means to note subtle changes in the strategy of accommodating challenges to locomotion. In unhampered walking, where no threat to stability is posed, both F and L are usually modulated almost equally to change velocity. Molen (4) noted this by referring to the s/n ratio where s was step length, and n was step frequency. Under circumstances where the walker perceives a threat to stability and intends to maintain velocity, the stride length is often shortened and the stride frequency increased. Applied to a younger person walking at "usual" speed, the minimum momentum of the body is large enough to assure safe movement through midstance despite any small disturbing impulse. An older person, on the other hand, whose "usual" velocity is slow, may perceive a threat to stability from a similar small disturbing impulse because of low momentum. That individual may: 1) shorten stride length and widen base of support; 2) increase stride frequency; and, 3) increase the double support period. These trade-offs were seen among all the subjects, particularly as they descended the ramp. The younger amputees, whose reaction times were presumed to be faster than those of the older subjects, demonstrated the lesser increase in F/L ratio. Subjects less sure of their security made pronounced increase in their F/L ratio in both ascending and descending the ramp.

SUMMARY

The effect on gait of five commonly prescribed ankle-foot devices was measured. The five devices were: SACH, SAFE, SEATTLE, ARTICULATED SINGLE AXIS, and MULTIPLE AXIS (Greissinger). Locomotion testing was conducted in a laboratory setting which provided testing on a level surface, surfaces with a lateral incline and decline, and up and down ramps. The principal objective measurements were angular acceleration of the prosthesis and forward velocity of the body.

Subjective ratings by the amputees served to resolve which physical variables determined the preference of a given ankle-foot device. The four physical variables were: 1) age; 2) ratio of actual to ideal weight; 3) relative length of residual limb; and, 4) the ratio of stride frequency to stride length. Thresholds were established which optimally classified devices preferred by the subjects.

Distinctions in performances were derived from analyses of the anterior-posterior angular accelerations occurring in the prosthesis immediately following heel strike. The accelerations were characterized as a damped oscillatory waveform. The principal factors serving to discriminate performances were amplitude and rate of damping. These findings, when related to the subjective rankings, showed that subjects preferred devices which developed the least shock and the greatest damping.

ACKNOWLEDGMENTS

We offer special thanks to the San Diego chapter of Amputees In Motion for their enthusiastic community effort in helping recruit subjects for the study. The authors gratefully acknowledge the efforts of Donald Johnson in the collection of test data and for the programming and data processing required in this project.

GLOSSARY

Correlation coefficient. A measurement used to express the degree of association between two variables.

Fourier coefficient. A number prefixed as a multiplier to a given sine or cosine component used in quantitating a periodic waveform. The magnitude denotes the relative contribution of that component to the total waveform.

Fourier series. The expansion of a periodic function into a series of sines and cosines.

Harmonic. One of the component frequencies characterizing a waveform and is an integral (whole number) multiple of the fundamental frequency. In gait, the fundamental is the stride frequency. Harmonics are integral multiples of the fundamental such as two times, three times, four times, etc.

Harmonic analysis. The analytical procedure of breaking a periodic function into components, each expressed as sine or cosine function and determining the relative contribution of each harmonic to the total waveform.

Harmonic ratio. The harmonic ratio is the quotient derived from dividing the sum of the absolute values of the even harmonics by the sum of the absolute values of the odd harmonics. Applied to this gait study, the higher the harmonic ration, the more symmetric the gait.

Impulse. Product of force times time in a linear system. Product of torque times time in a rotatory system.

Kinematics. Characterization of motions without consideration of what caused the motions.

Kinetic energy. Energy by virtue of a body in motion. Energy is the capacity to do work.

Kinetics. Consideration of causes of motion, hence the consideration of forces and energies.

Momentum. Product of mass times velocity in a linear system. Product of moment of inertia (for mass) about its axis of rotation times angular velocity.

Newton. The newton is the amount of force which imparts to a mass of one kilogram an acceleration of one meter per second per second. One newton is approximately 0.225 pounds.

Periodic waveform. A repetitive waveform, such as may be found to recur from gait cycle to gait cycle.

Potential energy. Energy by virtue of position. A raised weight has potential energy.

Regression analysis. A means to determine how two or more variables relate to each other.

Root mean square. The square root of the average of the sums of the squares of a variable. While the root mean square (rms) is almost the same as the arithmetic average, it allows for irregularities, in particular when numbers range above and below zero.

Significance level. Significance, also called the rejection level, is a value used to judge the probability of two variables not relating to each other. The symbol used to express the probability is "p." The smaller the "p," the less likelihood that the relation is pure chance.

Waveform. Shape of a wave as a function of an independent variable such as time, percent gait cycle, or other convenient base.

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Compressive strength mapping of femoral head trabecular bone

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Editor's Note

The following comments of a *JRRD* Editorial Board reviewer are presented to assist the reader in placing the merits of this work into perspective with regard to the methodology used.

The subject of this paper is of clinical interest in that surface replacement arthroplasty of the hip depends on the structural and mechanical integrity of the remaining portion of the femoral head following its surgical manipulation to permit appropriate fit of the surface cup.

"Cup Arthroplasty" has been performed in many thousands of pathological hip states including the arthritides, trauma, post-infection, congenital deformity, and others. A large clinical experience over several decades indicates that certain physical changes take place in the underlying bone structure due to mechanical and biological response to stress resulting from the cup surface, time related.

The *in vitro* mechanical research carried out on the osteoarthritic and normal femoral heads in this study has failed to take into consideration the large number of variables the biological material presents. For example, no histological information is included: the classification of specimens is superficial and vague. The pathological changes taking place in the subchondral bone of the femoral head with the diagnosis of osteoarthritis vary all the way from pathological cyst formation to extensive localized increased bone density (sclerosis) and are also dependent on a variety of physical and biological factors.

The paper has some confirmatory value pointing out that the hips described as having osteoarthritis present structural reproducible features differing from the so called normal hips. This is an obvious radiographic and clinical observation. Confirmation by objective testing, using the authors' methods, does quantify in these few specimens this variant. This material in print may be useful to others studying bone response.

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Abstract—The purpose of this study was to develop a test protocol and complete a relative three-dimensional mapping of the local trabecular bone compressive strengths in healthy and osteoarthritic femoral heads. Five fresh frozen cadaveric femoral heads were compared to five fresh frozen osteoarthritic femoral heads obtained from patients undergoing total hip replacement. Each femoral head was coronally sectioned into eight 5 mm slices, which were compression-tested at 20 sites using a hydraulic testing machine. The results of the femoral head compressive strengths were normalized, mapped, and evaluated. The test protocol produced reproducible results and allowed the determination of diseased portions of osteoarthritic bone. The regions of high compressive strength were similar for normal and diseased bone and were located in the superior medial

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portions of the head. Ultimate stress was higher in diseased bone than in normal bone, while the yield point and stiffness of diseased bone was lower than for healthy bone.

Key words: *biomechanics, femoral head, osteoarthritis, strength mapping, total hip replacement, trabecular bone.*

INTRODUCTION

Replacement of diseased hip joints with conventional total hip prostheses secured with polymethylmethacrylate (PMMA) is a successful procedure in the elderly, for whom the technique was originally developed. However, for younger patients (less than 60 years old), this procedure shows less promise. Various reports indicate higher rates of failure in the form of loosening and stem fracture than is the case in the elderly patients (3,11). Possible reasons for this higher rate of failure include: 1) higher levels of cyclic stress as a consequence of more frequent and strenuous physical activity; and, 2) redistribution of surrounding bone of the aging and/or changes in prevailing stress related to redistribution of loads by the implant (4,11,12).

Surface replacement was designed to replace only the diseased surfaces of the hip joint, while preserving the normal anatomy and biomechanical function of the joint and proximal femur. In comparison to conventional total hip arthroplasty, the proposed advantages of surface replacement include preserving normal bone stock, maintaining somewhat normal physiological bone loading patters, and decreasing the rate of infection. Surface replacement also allows for an easier method for replacement of failed implants.

These advantages have been offset in practice by high failure rates. The major failure mode is loosening of the femoral and/or acetabular cups. The reasons and contributing factors relating to these high failure rates include: crack propagation from areas of stress concentration along the rim of the implant, bone necrosis secondary to surgical disruption of the retinacular blood supply, poor instrumentation, bone remodeling in response to stress redistribution or disease processes, and finally, poor initial bone strength (3). This study was specifically designed to increase our understanding of the poor performance of surface replacement hip arthroplasty by evaluating one of these contributing factors—the initial bone strength of the femoral head.

As a result of the normal biomechanical function of the hip joint, the trabecular bone supporting a femoral surface replacement component is subjected frequently to compressive loading. If the mechanical strength of the underlying bone is not sufficient, this type of loading will result in segmental collapse or crushing of the trabecular framework and eventual loosening of the surface replacement component.

The extent of bony support in the diseased hip will depend directly upon the local magnitude and frequency of loading, as well as upon the location and extent of decreased bone integrity. Two of the mechanical properties of bone which may be indicators of bone integrity are the elastic modulus and ultimate stress associated with the crushing of the trabecular lattice.

The purpose of this study was to develop a test protocol to complete a relative mapping of the local trabecular bone compressive strengths of the femoral head. This mapping will allow a relative mapping of the compressive strength of the trabecular bone as it pertains to surface replacement hip arthroplasty, and will ultimately assist in the design of an improved surface replacement component.

MATERIALS AND METHODS

Five healthy fresh frozen femoral heads were obtained from male human cadavers over 60 years of age with no known history of hip disease. Five diseased femoral heads, removed during conventional total hip arthroplasty and fresh-frozen, were obtained from male patients over 60 years of age suffering from osteoarthritis. Patients were classified as osteoarthritic on the basis of a clinical history of degenerative joint disease and the gross appearance of bone in the hip joint in standard X-ray films. No attempt was made to sub-classify the osteoarthritis further. We wished to see if femoral head bone grossly classifiable as arthritic was different in gross compressive mechanical properties from the bone in normal femoral heads.

Each specimen was prepared by sectioning the femoral head along the coronal plane into six 5 mm-thick slices and two variable thickness end pieces, using a custom-designed bone saw. The coronal plane was selected because it would then be possible to test and map directly the change in compressive strength with distance below the femoral surface at any medial/lateral position, thus giving a sense of femoral head strength in sections which have the same orientation as common anteroposterior X-ray films of the hip. Also, the direction of testing was not considered crucial for two reasons. First, the trabeculae fail as much or more by buckling as they do by direct compression, and second,

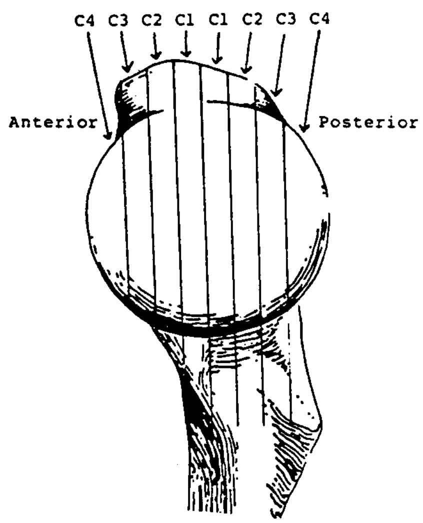

Figure 1a.
Illustration of femur, demonstrating the eight sections into which the femoral head was cut for testing.

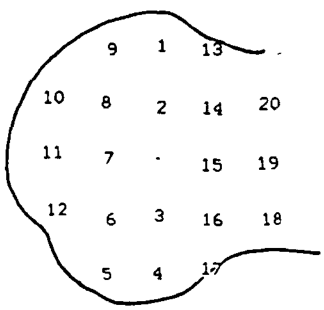

Figure 1b.
Femoral head test point grid.

the actual direction of femoral head trabecular loading *in vivo* varies substantially with joint motion.

An initial cut was made at the center of the femoral head, as measured from anterior to posterior. All additional cuts were referenced from the center cut, resulting in symmetrical anterior and posterior femoral slices. The slices were labeled as either anterior or posterior and as C-1, C-2, C-3, or C-4 (**Figure 1a**).

The center slices of the sectioned femur were marked with two points of reference. The first point, located with a compass, was the circular center of the weight-bearing surface of the femoral head. The second reference was the point of transition between the head and neck of the femur along its inferior surface. Using the center slices as an anatomical reference, a 2-mm hole was drilled through the remaining slices perpendicular to the cut surface through the circular center. This allowed the femoral slices to be bolted to a testing jig. Specimens were kept moist with normal saline throughout preparation and testing.

The compressive strength tests were performed with an Instron 1331 servo-hydraulic mechanical testing machine. Each femoral section was mounted to a custom-built indexing jig using a 2-mm bolt placed through the circular center. The rotational position of the femoral slice was fixed using the second reference point. The jig and secured femoral slice were positioned so that the Instron compression test rod was exactly over the circular center of the femoral slice. In this position, the jig established a square grid of testing sites on 10-mm centers across the surface of the femoral slice (**Figure 1b**).

Based on previous experience in our laboratory each test consisted of applying a compressive load perpendicular to the cut face of the slice (13). The load was applied at a constant rate of 0.05 mm/sec, using a 5-mm diameter cylindrical testing rod with the ordinary minimum edge radius necessary to remove machining burrs and give a clean edge. The resulting curves were plotted graphically by the Instron as load versus deformation. The ultimate strength, measured in newtons, was determined to be the maximum load on the load-deformation curve, and was converted to ultimate stress using the area of the testing rod. The full cross-sectional area of the rod was used in the stress calculations since the depth of rod penetration into bone at failure was always such that edge radius was well below the original bone surface. The elastic modulus was measured using the angle which the linear portion of the load-deformation curve made with the deformation axis. The slope was calculated as the tangent of the angle and then converted into appropriate units. Although both ultimate stress and elastic modulus were determined, ultimate stress was found to provide a more sensitive measurement than elastic modulus, and was therefore the principle value used in this study.

The ultimate stress results were averaged on a point-by-point basis for healthy and osteoarthritic bone and are presented in **Table 1** and **Table 2**. The ultimate stress results were then normalized for each femur by reporting all values as a percentage of the maximum ultimate stress measured for that femur. Normalization was necessary because the absolute strength of bone varied enormously from femur to femur. With normalization, it would be easier to see whether the relative strength changed in a consistent way with anatomic position and whether the pattern was different for normal and osteoarthritic bone. These normalized results were then averaged for the five samples of healthy and osteoarthritic femoral heads and represented

both quantitatively in **Table 3** and **Table 4**, and symbolically using circles of varying diameter. A circle of larger diameter indicates a trabecular bone test site with higher ultimate stress to failure (See **Figure 2**).

The reproducibility of the testing technique was evaluated using a paired comparison of the average normalized compressive strengths at each of the 20 adjacent test points of the central two femoral slices C-1 anterior and posterior for all 10 specimens. Reproducibility would be indicated if, on a point-by-point basis, the average standard error was small and the correlation coefficient for a least squared regression of the paired points was high, since the slices and corresponding points were adjacent. The data were also examined to see if results for normal and osteoarthritic bone were qualitatively or quantitatively different.

RESULTS

The paired comparison of the average compressive strengths of the 20 adjacent test points on the two central slices resulted in an average standard error on a point-by-point basis of 1.15 percent, with a standard deviation of 7.29 percent. A least squares regression conducted on the same data points resulted in a slope of 0.91, with a correlation coefficient of 0.94.

The results of the average ultimate compressive strengths (in MPa) of five femoral heads on a point-by-point basis are presented in **Tables 1** and **2** for both healthy and osteoarthritic bone. The anterior and posterior sections C1 to C4 refer to the femoral head slices shown in **Figure 1a**, while points 1 to 20 refer to the test point grid shown in **Figure 1b**. At test point 2 on anterior slice C1, for example, the average stress of the five femurs was 18.8 MPa for the healthy femurs, and 11.2 MPa for the diseased femurs.

The results of the averages of the normalized ultimate compressive strengths for each femur are presented on a point-by-point basis in **Table 3** and **Table 4**. These values are felt to more accurately represent a relative strength mapping for femoral heads as each test point in a given femur is normalized, using the maximum ultimate compressive stress for that femur. Using **Table 3** as an example, the ultimate stress of test point 2 in anterior slice C1 aver-

Table 1.
Healthy femoral heads. Averages on a point-by-point basis of the ultimate compressive strengths of the 5 healthy femurs. (Results in MPa.)

	Anterior				Posterior			
Point	**C4**	**C3**	**C2**	**C1**	**C1**	**C2**	**C3**	**C4**
1		17.6	12.0	12.5	10.8	11.3	8.0	
2	12.1	14.4	20.9	18.8	16.6	16.8	9.8	6.0
3	5.4	5.3	4.5	5.1	5 9	4.8	4.1	7.1
4		4.3	1.0	2.1	5.8	7.7	7.7	
5		12.9	6.3	12.2	12.2	16.4	8.1	
6	15.3	8.6	8.3	6.2	6.1	9.5	11.1	15.0
7	14.0	11.7	15.1	14.8	17.7	14.6	16.3	23.7
8	20.5	16.1	19.8	20.8	22.0	14.6	21.4	19.4
9		9.3	24.0	22.1	17.3	16.2	14.1	
10		5.9	19.1	21.4	18.1	14.6	21.4	
11		6.9	16.9	10.6	10.8	12.3	22.8	
12		6.9	12.7	8.6	9.1	9.9	17.8	
13		11.1	8.7	5.0	4.6	3.5		
14	4.6	6.9	4.6	4.9	5.0	2.9	2.2	0.2
15	2.4	5.5	10.3	12.6	10.3	7.7	4.1	3.3
16	1.5	3.4	8.7	11.6	9.6	10.6	6.1	0.5
17			7.0	7.7	11.6	9.1		
18		4.3	9.2	10.7	9.1	9.0	5.5	
19		4.7	5.2	2.5	2.4	4.3	2.3	
20		6.0	5.0	4.7	3.2	6.4	2.0	

Table 2.
Diseased femoral heads. Averages on a point-by-point basis of the ultimate compressive strengths of the 5 diseased femurs. (Results in MPa.)

	Anterior				Posterior			
Point	**C4**	**C3**	**C2**	**C1**	**C1**	**C2**	**C3**	**C4**
1		9.2	7.1	9.8	9.8	8.6	11.1	
2	14.7	12.9	12.7	11.2	10.4	10.3	9.5	5.6
3	11.5	9.3	10.7	10.5	13.0	10.4	7.5	3.7
4		4.5	4.2	6.5	8.7	2.4	10.3	
5		9.9	6.3	5.8	4.7	11.3	13.8	
6	11.7	9.1	8.3	7.8	8.5	8.5	5.4	7.2
7	16.4	16.4	23.2	21.4	19.1	16.8	15.0	14.0
8	30.3	25.0	19.3	21.5	22.7	17.2	14.3	22.3
9		25.0	26.8	27.7	21.3	27.2	9.5	
10		22.8	26.0	24.7	30.1	25.6		
11	8.3	19.2	15.3	16.8	21.1	19.4	16.0	
12		11.8	13.7	11.5	13.3	10.3	12.7	
13		8.5	11.0	8.4	8.1	7.9		
14	25.2	8.4	3.5	7.7	5.6	4.7	2.7	1.3
15	19.3	9.6	13.7	14.2	16.5	9.3	4.9	3.9
16	17.3	16.6	11.5	14.0	17.2	15.9	6.2	
17								
18			6.0	4.[illegible]	7.4	8.2		
19		4.9	4.7	10.6	7.4	1.2		
20		4.1	5.6	5.7	7.4	2.7		

aged 41 percent of the maximum stress of the femur.

The results of the compressive strength mapping of femoral head trabecular bone for both healthy and diseased bone is shown graphically in **Figure 2**. Again, the circles of larger diameter represent higher average ultimate compressive strengths. This compressive strength map shows the strongest bone to be in the superior medial portion of the femoral head for both normal and diseased bone.

For all slices, the range of maximum test rod indentation at failure was between 0.1 and 0.7 mm. The average ultimate compressive strength for all test points of the healthy femur was 9.98 MPa (SD = 6.21 MPa), while the compressive strength of the osteoarthritic femur was 12.2 MPa (SD = 6.72). A comparison of the average compressive strengths for the osteoarthritic and healthy femurs using a pooled variance *t*-test indicated that the difference was statistically significant with $p < 0.05$.

In addition, the ultimate stress was found to be linearly proportional to the elastic modulus. A regression plot is not presented because of the large amount of data analyzed (10 femoral heads × 120 tests per head = 1200 values). A least squared error regression analysis indicated that this relationship was significant with $p < 0.10$, with a correlation coefficient of 0.09. The relationship is as follows: Ultimate Stress = (0.035) × Elastic Modulus.

DISCUSSION

The compressive properties of trabecular bone have been determined in several studies by testing uniformly shaped specimens obtained from whole bones (1,5,6,7,8, 12,14). The main disadvantage to testing cut specimens is that cutting may damage the specimen and change the mechanical properties of the specimen.

The compressive strength test protocol developed in this study was based on the following criteria: first, the test should be performed essentially *in situ*, with as much surrounding bone in place as possible. This would reduce the risk of damaging the specimen or disturbing its interactions with the surrounding environment. Second, in order to correlate the information from compressive testing between specimens, the site of each test must be referenced to fixed landmarks. Finally, the technique should be

Table 3.
Healthy femoral heads. Averages on a point-by-point basis of the normalized ultimate compressive strengths of the 5 healthy femurs. (Results in percentages.)

	Anterior				Posterior			
Point	C4	C3	C2	C1	C1	C2	C3	C4
1		58	41	41	36	36	27	
2	39	47	72	65	57	53	32	20
3	18	18	16	17	20	16	13	22
4		18	3	7	19	25	27	
5		46	22	41	38	53	27	
6	51	30	30	22	21	34	39	50
7	47	39	49	51	60	50	56	79
8	68	55	65	68	71	48	72	63
9		29	78	73	56	53	41	
10		23	62	69	60	49	64	
11		26	53	35	36	42	74	
12		24	41	28	29	34	54	
13		36	31	18	17	12		
14	16	23	17	17	17	10	7	1
15	8	17	35	42	36	27	14	10
16	5	11	28	40	33	38	21	2
17			28	28	42	32		
18		11	31	37	32	32	17	
19		15	17	8	8	14	8	
20		19	16	15	11	21	7	

Table 4.
Diseased femoral heads. Averages on a point-by-point basis of the normalized ultimate compressive strengths of the 5 diseased femurs. (Results in percentages.)

	Anterior				Posterior			
Point	C4	C3	C2	C1	C1	C2	C3	C4
1		26	18	24	24	20	25	
2	38	31	31	28	26	25	23	13
3	28	23	25	24	29	24	17	8
4		10	10	15	20	7	21	
5		21	16	13	11	28	28	
6	30	23	20	19	20	20	13	16
7	41	41	57	51	45	39	35	33
8	69	58	46	53	57	43	34	51
9		64	62	62	48	62	28	
10		60	61	58	68	60		
11	24	45	36	40	49	44	38	
12		24	32	29	32	25	26	
13		22	26	21	20	20		
14	63	20	9	18	14	12	6	3
15	49	23	32	35	39	22	13	8
16	43	43	26	33	39	36	15	
17								
18			14	12	20	16		
19		14	11	24	18	3		
20		12	13	14	18	8		

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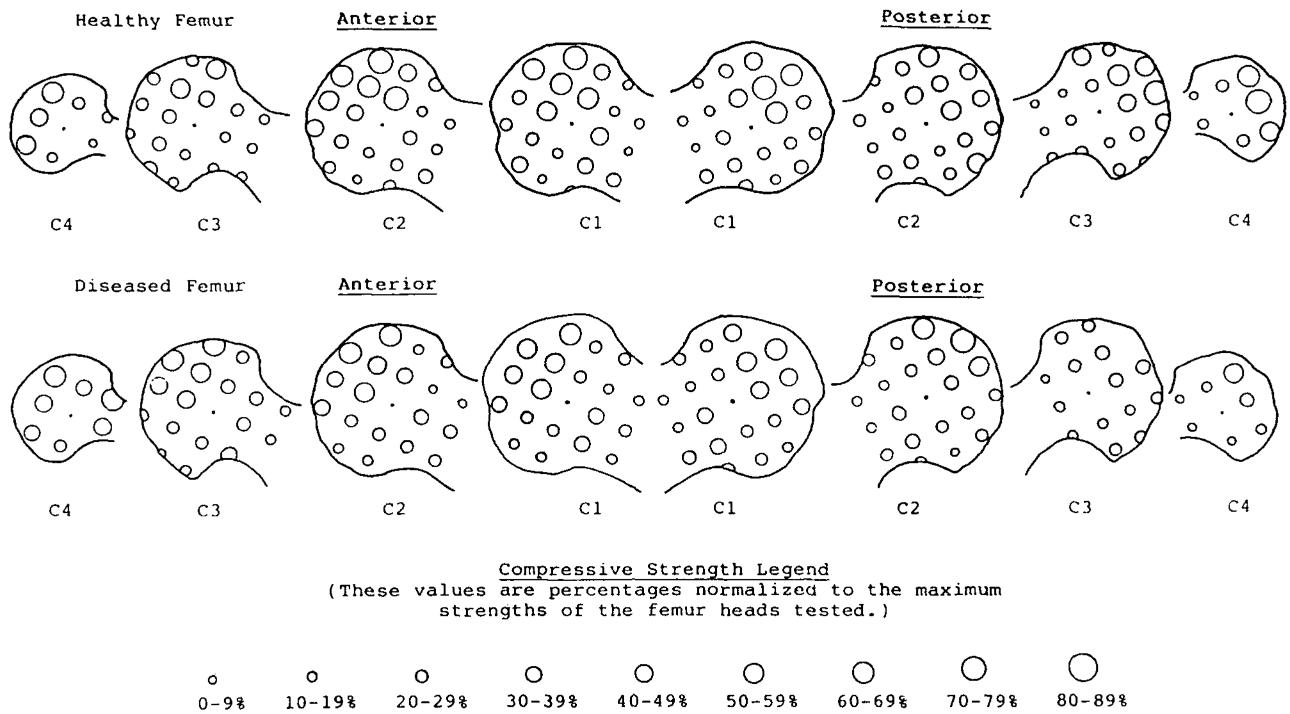

Figure 2.
Femur compressive strength averages (normalized results). These values are percentages normalized to the maximum strengths of the femur heads tested.

reproducible and accurate. It was felt that the technique described previously would meet a majority of the above criteria. Special care was taken in the selection of a testing rod and test site grid because of their importance in obtaining valid data.

The 5-mm diameter rod was selected as a compromise. A smaller test rod would be more prone to measure the strength of individual trabeculae rather than an average trabecular lattice strength, while a larger rod would not allow enough test sites on each femoral slice to produce a compressive strength map with useful detail.

The 10-mm center-to-center test grid was established to maximize the total number of possible test sites while avoiding error due to test area overlap. This was evaluated by testing-to-failure a number of sites on a femoral head slice, using a 5-mm rod. Radiographic evaluation of cross-cut sections of the test sites revealed plastic deformation of the trabecular bone up to 1.5 mm from the outer edge of the test rod. Combined with a test rod radius of 2.5 mm, this resulted in a total radius of potentially damaged bone of up to 4 mm from each test site. A test grid on 10-mm centers leaves a 2-mm buffer zone of undamaged bone between test sites.

To evaluate test protocol reliability, a paired comparison was conducted on the central two slices. Since all of the 20 test sites were adjacent *in vivo*, each test site should have the same compressive strength as its corresponding pair. Thus, in a paired comparison, the average of the differences between paired test sites should approach zero. In addition, a least squares regression of these paired data should result in a line with a slope of 1.0. The mean standard error on a point-by-point comparison was 1.15 percent with a standard deviation of 7.29 percent. The linear regression resulted in a slope of 0.91 with a correlation coefficient of 0.95. Thus, the selected technique was felt to be reliable.

Figure 3 demonstrates that areas of diseased bone within each osteoarthritic femoral head slice could be identified by visual inspection of the shapes of the test curves. As shown, curves for healthy bone are smooth in shape and similar in slope. While most of the test sites in an osteoarthritic femoral head displayed curves similar to those of normal femurs, specific areas within each osteoarthritic femoral head displayed curves of either dramatically lower modulus or irregular shape. Furthermore, the mode of failure observed from these curves for the normal

femoral head trabecular bone and much of the osteoarthritic bone appeared to be abrupt and catastrophic, probably representing an acute collapse of the underlying trabecular lattice. For the diseased portions of osteoarthritic bone, however, the failure mode appeared to be segmental with the trabecular bone collapsing in stages.

Goldstein reported that the ultimate strength of trabecular bone in the proximal tibia is linearly proportional to the elastic modulus, and demonstrated the same patterns of variability (6,7). The results of this study also indicate that a linear relationship between ultimate stress and elastic modulus exists: Ultimate Stress = 0.035 × Elastic Modulus. The constant for the above equation (0.035) is 17 percent higher than that calculated by Goldstein (0.028) (7).

The maximum strengths were precisely measured in most cases due to the obvious point of inflection of the load-deformation curve. The elastic modulus, on the other hand, was less precisely measured. For the strain rate used in this study, the majority of the elastic modulus angles ranged between 82 and 86 degrees with a measured accuracy of only 0.5 degrees. Thus, a relatively large change in ultimate strength was required before a corresponding change in elastic modulus could be measured. Since ultimate strength and elastic modulus are not independent variables, and because of a higher measured precision, ultimate strength values were used for the three-dimensional mapping.

The ultimate strength was difficult to determine for the diseased portions of osteoarthritic bone, due to the irregular curves which were produced (**Figure 3**). The ultimate strength was measured at the first point of inflection noted in the rising load-deformation curve. As was mentioned previously, this type of failure was likely due to segmental collapse rather than simultaneous lattice failure, thus providing a less precise ultimate strength measurement. Therefore, the elastic modulus might have provided a more precise measurement for these curves.

For the diseased portions of osteoarthritic bone, the elastic modulus was, on the average, 20 percent lower than would be anticipated from the ultimate strength measurement. While not statistically significant ($p > 0.05$), the change was consistent enough to indicate that the diseased portion of osteoarthritic bone has a lower elastic modulus than equivalent healthy bone. The fact that the relationship between elastic modulus and strength was different for bone from osteoarthritic and nonosteoarthritic patients suggests that the two types of bone are different in their structure in some way other than density, due to differences in porosity. This certainly seems possible, since from an engineering view, bone is a composite material with a complex structure. For example, changes in the relative amount of mineral and collagen constituents would alter the strength/stiffness relationship. However, we did not attempt any chemical or histologic analysis to verify this hypothesis.

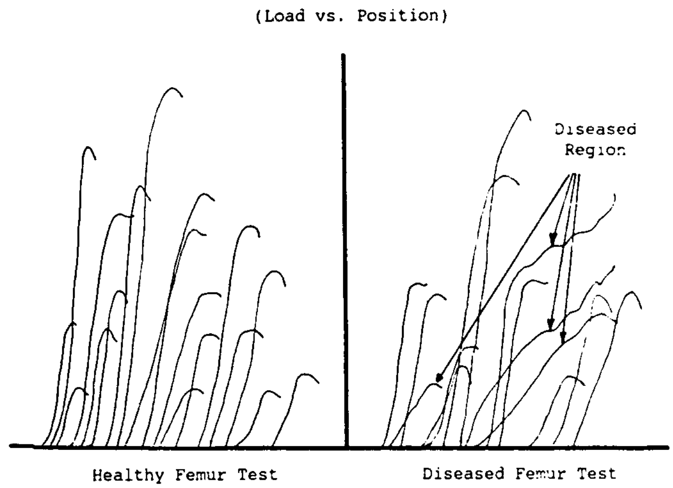

Figure 3.
Samples of Instron compression tests on femurs.

A comparison of the compressive strength mapping of healthy and osteoarthritic femoral head trabecular bone shown in **Figure 2** revealed that the regions of high compressive strength (ultimate stress) were the same for both normal and diseased bone, and were located in the superior medial portions of the head. Thus, no consistent change in the ultimate stress distribution was found between healthy and diseased bone.

The ultimate stress was higher on the average in osteoarthritic bone than in healthy bone. The average ultimate strength for the normal and diseased bone were 9.98 MPa (SD = 6.21) and 12.2 MPa (SD = 6.72), respectively. A pooled variance *t*-test was performed on a point-to-point basis, using the average normalized strength values for equivalent points. The result showed that the strength of the two types of bone was significantly different ($p < 0.05$). These mean values are within the range of trabecular bone compressive strengths obtained by other investigators using other methods (2,5,12,14). The large standard deviations are likely due to normal biologic variation in bone properties between individuals, combined with experimental error produced by too small a test rod diameter and test site variability.

The etiology behind a higher average ultimate stress

in osteoarthritic bone is not clear. One possibility is that segmental failure, seen in the diseased portions of osteoarthritic bone, may cause falsely elevated ultimate strength measurements as the interstices fill up with trabecular fragments during failure, which increases the bone density and resists further penetration. Coronal plane testing, however, should provide less of a problem with trabecular fragments.

Another possibility is that clinically, in patients with osteoarthritis, the lack of articular cartilage leads to bone deformation associated with a more nonuniform load distribution. Certain portions of the bone are therefore required to support a larger percentage of the load. This bone may subsequently remodel to accommodate the increased load resulting in areas of bone with higher ultimate strengths (9). Finally, osteoarthritis is thought to begin as numerous microfractures within the trabecular lattice. The reparative process involved in this condition may result in the formation of specific areas of dense bone, thus increasing the average ultimate strength (10).

It is clear from clinical results to date that substantial improvements must be made in the technology of surface replacement hip arthroplasty before it will be widely used. The information from this study may be helpful in *understanding surface replacement hip arthroplasty* failure modes and devising methods to improve this approach to hip replacement.

CONCLUSIONS

- The compressive strength testing protocol outlined in this study is reproducible, and allows the determination of diseased portions of osteoarthritic trabecular bone.
- The failure mode for the normal trabecular bone and much of the osteoarthritic bone was abrupt, representing an acute collapse of the underlying trabecular lattice.
- The failure mode for the diseased portions of osteoarthritic bone was segmental, with the trabecular bone collapsing in stages.
- The elastic modulus of the diseased portions of osteoarthritic trabecular bone was on the average 20 percent lower than corresponding healthy bone.
- For trabecular bone there exists a linear relationship between ultimate stress and elastic modulus: Ultimate stress = 0.035 × Elastic Modulus.
- The regions of high ultimate strength were the same for both normal and osteoarthritic trabecular bone and were located in the superior medial portions of the femoral heal.
- The ultimate strength was higher on the average for osteoarthritic bone than in normal bone.

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Patient and staff acceptance of robotic technology in occupational therapy: A pilot study

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Abstract—While the majority of applications of robotics in the field of rehabilitation focus on the development of smart aids for people without upper extremity function, there is also potential for the robot as a therapy "aide." We designed, built, and pilot-tested hardware and software that used a robot to provide muscle reeducation movement patterns after stroke. This is a report on a field trial, in which 11 occupational therapists used the system with 22 patients; each patient averaged 2.2 sessions. Based on information contained in the system database, a log, patient interviews, and therapist questionnaires, we evaluated safety, system utility, and patient and therapist acceptance. The results suggest that robotic treatment is safe and accepted (if not welcomed) by patients. The therapists expressed a qualified acceptance, suggesting several modifications to increase utility. The potential for the application of robotics in rehabilitation therapy is discussed in light of these findings.

Key words: *manpower, movement patterns, occupational therapy equipment, rehabilitation therapy, stroke.*

INTRODUCTION

Robots were first introduced into the field of rehabilitation as mechanical personal care attendants (PCAs), or smart aids. Currently, at least nine different groups in five countries are at work on further development of what is variously called a robotic aid, manipulator, robot arm, or work station (4-6).[1-9] These projects have different premises and goals. The robots are in different stages of development; the only one being marketed commercially (as of this writing) is the Boeing workstation,[3] now known as the PRAB Command 1, a voice-activated personal workstation that makes it possible for a quadriplegic person to perform many clerical and managerial jobs. The Johns Hopkins robot arm (with chin control rather than voice input) can brush teeth and feed.[4] The Palo Alto VA/Stanford robotic aid can do selected vocational tasks and activities of daily living. Other features under development are vision, in the Spartacus project,[5] and mobility, in the Palo Alto VA/Stanford project.[7]

It is likely that in the future rehabilitation robots will be mobile, have vision systems and speech input/output, and will perform a large number of tasks in the home and workplace. Whether they will be affordable and acceptable to people with a disability is yet to be seen. Early evaluation results seem to indicate a positive response (5).[6]

Robot as therapy aide

This project has taken robotics technology in an alternate direction: instead of the robot as PCA, we have worked toward the robot as occupational therapy aide. A search of the literature has shown that this application is rather unique. A recent introduction to robotics and the disabled published in an occupational therapy journal even fails to mention the concept of therapeutic applications (1). Engelhardt (3) lists a number of *possible* uses of robotics in health care settings, including therapeutic ones. Therapeutic applications were actually developed by Khalili and Zomlefer (7) who constructed a continuous passive motion

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robot; the Cambridge group which built a manipulator to assist in the developmental education of young children with severe physical impairments (6), and Engelhardt and colleagues, who piloted robots for range of motion of wrists and ankles.* However, it appears that none of these applications has gone as far as ours in using the ability of robots to "sense," "think" and "act."

The opportunity for cost savings was one incentive behind our project, which aimed to explore the possible uses of robotic technology in rehabilitation therapy. With demands from third party payors for increased contact time and performance improvement at lower cost and in shorter time, rehabilitation facilities must find innovative means for delivering quality therapeutic treatment without increasing staffs or budgets. A second incentive was improvement in therapy. Even if the cost savings realized are minimal, a robotic system has possible advantages. It can repeat the same movements many times with a high level of precision. Its capacity to count and to time patient activity in great detail enables it to produce objective and detailed reports of patient performance. The objective monitoring of patient progress and the pinpointing of problem areas will contribute to quality therapy.

The aim of this report is to describe the results of a pilot study which explored the safety, utility, and acceptance of one particular robotic-system module designed for stroke movement patterns therapy.

Robotic system: Design and application

Background. Since stroke patients are a large part of the population in any rehabilitation setting, this group was selected for the pilot study, which focused on motor recovery. The stages of muscle reeducation after a stroke focus first on normalizing tone, followed by facilitating basic mass functional movements, and progressing to more isolated advanced functional patterns.

To implement movement patterns in the early stages of recovery, the therapist may move the affected extremity through a variety of therapeutic patterns, while facilitating appropriate muscle responses by stroking, tapping, holding, etc. In later stages, therapist assistance is decreased as the patient begins to move his limb more independently. Patterns increase in complexity as the patient progresses. In the occupational therapy clinic, activities to facilitate these movement patterns may include picking up, transferring, or touching objects at specific points in space as directed by the therapist. The points vary in direction and height depending on individual needs. This treatment has two drawbacks: 1) intensive muscle reeducation requires one-on-one treatment which is limited by time restrictions and staffing issues; and, 2) the therapist has no means of collecting quantitative data on the nature and frequency of the patterns and on patient performance. Although the latter is not critical, the capacity to do so would help in duplicating treatments for consistency, in providing justification for treatment, and in verifying progress in an objective way.

The research team decided to develop a robot system to aid the therapist in providing upper extremity reeducation for stroke patients. While the technical requirements to do so would be less complicated than, for example, writing a robot toothbrush program, it would put the applicability of robotics in rehabilitation therapy to an acid test. We wanted to demonstrate that stroke patients, even those with diminished cognitive competence and no familiarity with automated equipment, could be administered therapy safely and show patient acceptance of this technology; the results would encourage a variety of applications in occupational and physical therapy.

Design. The robot system and specialized hardware and software components were designed and developed by a research team consisting of occupational therapists at the Rehabilitation Institute of Michigan, Detroit, MI, and engineers at Metropolitan Center for High Technology, Detroit, MI. It was tested in the occupational therapy clinic of the Rehabilitation Institute of Michigan (2).

The system (**Figure 1**) consists of the UMI RTX robotic arm controlled from an IBM-PC personal computer. This arm has six degrees-of-freedom, or six axes of jointed movement. These joints duplicate the range of possible vertical and horizontal movements which a therapist would use with a patient within a 3-foot work envelope. The RTX arm has specific safety features which make it an ideal choice in personal robot applications—it moves slowly and can be stopped by a moderate touch.

Two sensor switches with indicator lights monitor the patient's movements. A "target" switch is mounted on the robot arm gripper or "hand." A "home" switch is placed either on the patient's lap or on a stool at the patient's side. A custom-built data acquisition board within the computer collects the switch sensor information. Computer programs were written to control the robotic arm exercise movements, collect the patient demographic and exercise data, and generate patient performance reports using the computer's printer.

The exercise procedure is available in two versions:

*Personal communication with K.G. Engelhardt, 1988.

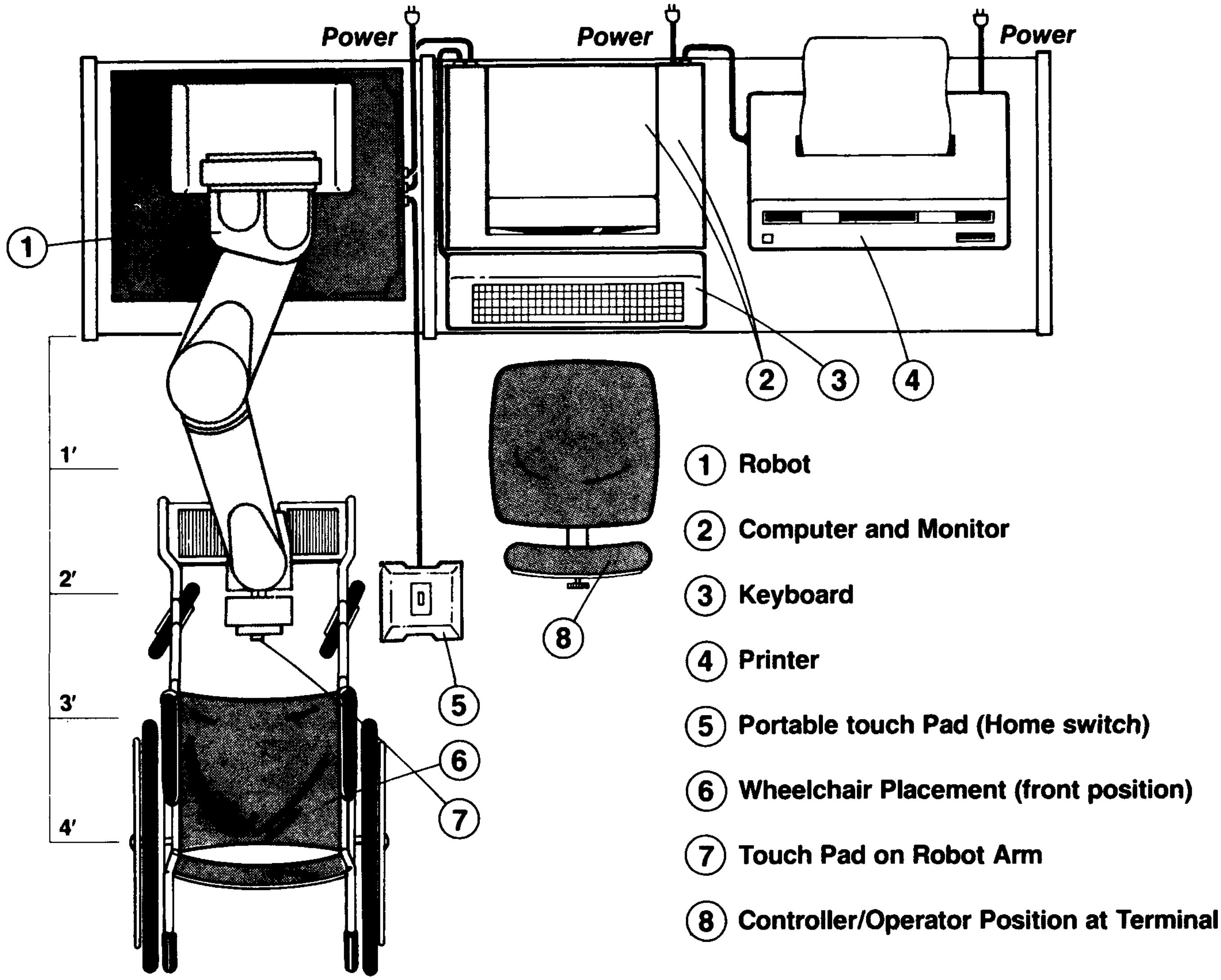

Figure 1.
Components of the robotic system.

the Pace mode and the Wait mode. The Pace mode requires the patient to work at one of four predetermined speeds, while the Wait mode allows him to work at his own rate. Five movement patterns, each consisting of eight points in space, are preprogrammed. These patterns have applications for various patient problems; they vary in difficulty, choice of points in space (left, right, low, high), and the sitting balance required. During a pattern run, each point is visited three times for a total of 24 exertions by the patient. An exercise routine with the patient seated in his wheelchair facing the robot arm (**Figure 2a** and **Figure 2b**) proceeds as follows:

- The indicator light on the "home" switch cues the patient to touch it. Once the patient has touched the switch, the light goes off.
- The robot arm moves to a point in space, then stops. The "target" switch light cues the patient to touch the robot arm gripper switch. (In the Wait mode the "target" switch light remains on until the patient touches it; in the Pace mode the light remains on for a predetermined amount of time depending on the speed chosen. If the patient fails to touch the target switch within that time, a "miss" is recorded and the home switch light goes on again).

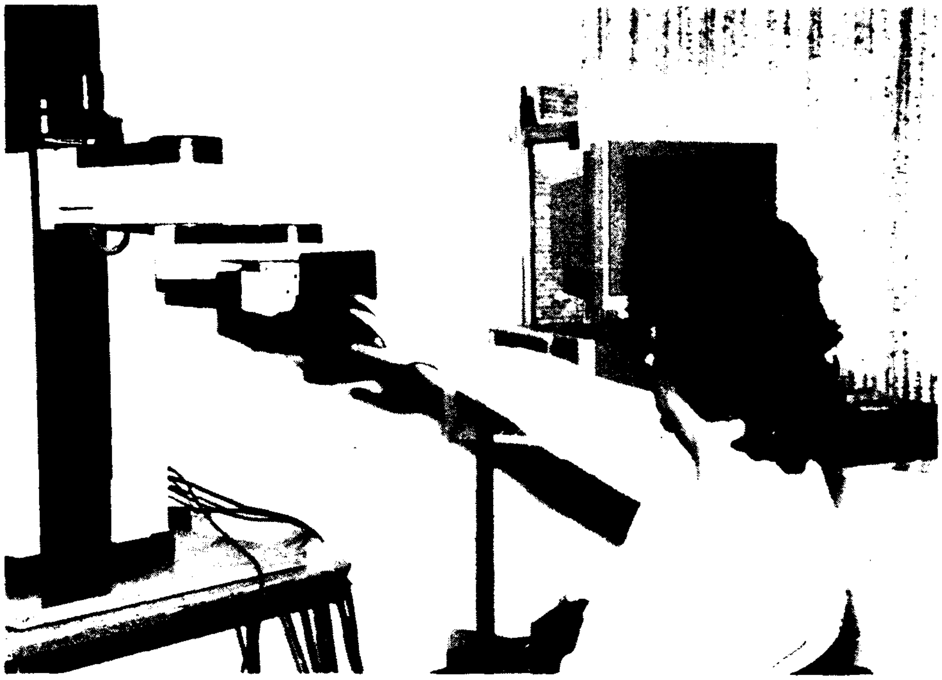

Figure 2a.
Patient using the system by touching the pad on the end-effector.

Figure 2b.
Patient using the system by touching the pad placed on the floor.

- The patient then touches the "home" switch again.
- The robot arm moves to the next position.

This cyclic procedure is repeated until the exercise routine is completed. The indicator lights and beeper give the patient visual and auditory feedback. The repetitive reaching and touching results in the patient performing the same pattern movements traditionally performed with a therapist.

A typical session involves the therapist entering information on the stroke patient into the computer, such as name, ID number, and Brunnstrom stage of recovery of the arm (9). The therapist then selects the appropriate exercise pattern, mode, and speed. After the patient has completed the cycle for the selected pattern and rested as necessary, the pattern may be repeated (at the same or higher speed), or the patient can be switched to a more difficult one. This decision may be based on feedback on the patient's performance provided by the system. At the end of a session, the printer can produce a report with essentially the same performance information for future reference (**Figure 3**). The system documents what specific treatments the patient has received: patterns, repetitions, speeds, and "hits" and "misses" (successful and unsuccessful attempts to touch the switches within the specified time), and the points in space where these occurred.

OBJECTIVES AND METHODS

During a 5-month period, the system was used in the occupational therapy clinic of the Rehabilitation Institute of Michigan as part of a pilot study conducted according to a protocol approved by the Institute's Institutional Review Board. The objectives of the study were to: 1) determine safety of the system for the patients; 2) assess acceptance of the system by the patients and the therapists; and, 3) explore utility of the robotic system as perceived by the therapists.

During the study, 11 therapists used the system with 22 patients. All the therapists were female, with an average age of 30 years. They averaged 7 years of experience in OT (range 0 to 17); only one had previous experience with computers. The therapists received inservice training in small groups (5 to 8 people) in order to become familiar with the aims of the project, hardware and software, operational skills needed, and use of the log. After that, supervision and consultation were provided as needed by all members of the research team, especially the research occupational therapists.

The patient group consisted of 10 females and 12 males; the average age was 53 (range 15 to 80). Eight were outpatients and 14 were inpatients. Diagnoses were recent stroke with right hemiplegia (N=8) or left hemiplegia (N=9); Guillain-Barré syndrome (N=1), traumatic brain injury (N=1), multiple sclerosis (N=2), and amputation (old stroke) (N=1).* The Brunnstrom stage of the affected hand of the stroke patients ranged from 1 to 6; the stage of the arm ranged from 1 to 5.

These patients had a total of 46 sessions with the system (average of 2.2 per patient), during which 70 cycles were completed, 38 in wait mode and 32 in pace mode. All the patterns and speeds were used. All patients and therapists in the study were volunteers; they received information on the purpose and significance of the study, and signed an informed consent document.

Information was collected from four sources: 1) a log located next to the computer in which therapists recorded comments, suggestions, and system problems; 2) the system database; 3) patient feedback forms, completed with help from the therapist (aphasic patients answered with nods) (**Appendix A**); and, 4) a comprehensive therapist questionnaire, completed at the end of the pilot study (**Appendix B**). Because of the small number of patient and therapist participants and the nature of the data, the presentation of results that follows is qualitative rather than quantitative.

*While the system was designed for use with stroke patients, therapists were free to explore its potential uses with other diagnostic groups. The therapists' opinions on safety, utility, etc. of the system reported here are based on use with *all* patients, whatever their diagnosis.

RESULTS

Safety

The safety of patients and therapist was an overriding concern of the research team. We used markers on the floor to indicate exactly where the wheelchair was to be placed in front of the arm for safety and maximum effectiveness. In performing the movement patterns, the patient's arms or hands were never within the area of space *actually used* by the robot: the sensor was mounted on the most distal point of the arm and needed touching only. No safety problems were mentioned in the data sources available.

Being safe is different from feeling safe: if a patient felt threatened in any way by the robot arm, it would have resulted in a refusal to use the system, or less-than-optimal cooperation. All 20 patients completing the form answered affirmatively to our question whether they felt safe (**Appendix B**, Q 4).

Staff were asked whether at any time they felt that their patient was at risk of being hit by the robot arm (**Appendix B**, Q 10). The answers indicate that such fears existed prior to the therapists becoming familiar with the system, or in the case of a patient who was very slow in moving back into his chair after touching the target switch. They also stated that they would have no problems letting cognitively intact patients work alone, but many specified that some form of supervision is needed for patients with decreased mentation or those who get easily confused. Three therapists thought that it was necessary that for these patients the therapist or a therapy aide should be at their side, but three others thought it sufficient to keep the patient in one's visual field.

Patient acceptance

The patients' acceptance of the robotic aide is summarized in **Table 1**. All therapists indicated that overall the patients' responses to the robot were positive. In their comments, they added that some patients thought it boring, but others thought it interesting. Many answers suggested that patients appreciated how the system gave them a "real workout."

System utility

Several questions on the therapists' questionnaire addressed issues of system utility. While therapists were

Table 1.
Patient Feedback: Acceptance*

Question	Answer yes	no
Did you like this treatment?	19	1
Do you feel it was helpful?	19	1
Was this boring?	4	16
Was it confusing to use?	4	16

*Based on answers by 20 patients: 18 first-time users, plus two who completed the feedback form at their last session.

learning the system, setting up a patient (including positioning the patient, turning on the equipment, identifying the patient and selecting pattern and speed) might take quite some time. Almost all became more adept. The information in **Table 2** on the therapists' estimate of the set-up time needed (after they had become familiar with the system), and their satisfaction with that time suggests that the maximum time set-up should take is 5 minutes. Reasons as to what took the longest in setting-up varied. Two therapists indicated that positioning the patient, explaining to the patient or turning on the equipment was the most difficult or time consuming, and seven named entering of the necessary information on themselves and their patients. Two people stated that both were time-consuming. The set-up process was judged not difficult by five therapists.

While data input may have been time consuming, most of the therapists had no particular problem. Ten stated that the instructions on the screen were easy to follow, and only three recommended specific changes that would make use of the system easier. Six therapists reported problems with the computer program and six reported malfunctioning of the robot arm.

Table 2.
Therapists' estimate of the time needed to set up, and their satisfaction with that time.

Estimated set-up time needed (minutes)	Satisfactory yes	no	total
10,15	2	1	3
5-10		2	2
5 or less	4		4
2-3	1		1
Not given		1	1
Total	7	4	11

Only four therapists reported that they actually used the performance information for making a decision as to what pattern and/or speed to select for the next cycle or session, or for reporting patient progress in chart rounds. Those who did not use the information gave various reasons, including: lack of understanding of what the data meant, loss of data due to computer (program) malfunction, and lack of applicability of the data to the problem of their particular patient.

The report format itself was called "clear" by seven subjects. One therapist suggested that exact range-of-motion be included. Eight therapists had produced the print report (**Figure 3**), and did not experience any problems. Printouts were made not just to have a permanent record allowing comparison over time; several therapists indicated that they showed the printout to the patients because they were most interested in their scores.

All 11 therapists stated that the five preprogrammed patterns were adequate for their needs, although three qualified that by stating that more complex patterns would be welcome. All stated that they were able to obtain therapeutic movements from the established patterns. Seven therapists expressed a preference for an option allowing programming of patterns by the therapist because custom-made patterns would better serve the patients' needs.

All therapists stated that the system was a valuable treatment modality for motor relearning. Reasons given included: 1) provided the repetition and consistency needed for relearning; 2) elicited patient cooperation; 3) delivered exactly the prescribed patterns every time; 4) provided therapeutic motions uni- and bilaterally; and, 5) helped with motor planning, praxis, and spontaneity of movement. They suggested that the system needed further development for use by patients with higher Brunnstrom levels. Others suggested developments that included functions for relearning fine motor aspects of hand function and for advanced coordination or more distal gross motor movements.

Even though the pattern program was designed specifically for stroke patients, the therapists saw applicability to a number of other diagnoses and used it with other diagnostic groups. Suggestions for potential users included traumatic brain injury patients who need upper extremity reeducation, patients with burns or peripheral neuropathy, multiple sclerosis, systemic lupus erythematosus, or arthritis, and patients with quadriplegia or polyneuropathy. The purpose of using the pattern program with these patients was not necessarily for motor relearning; therapeutic exercise for improving range of motion,

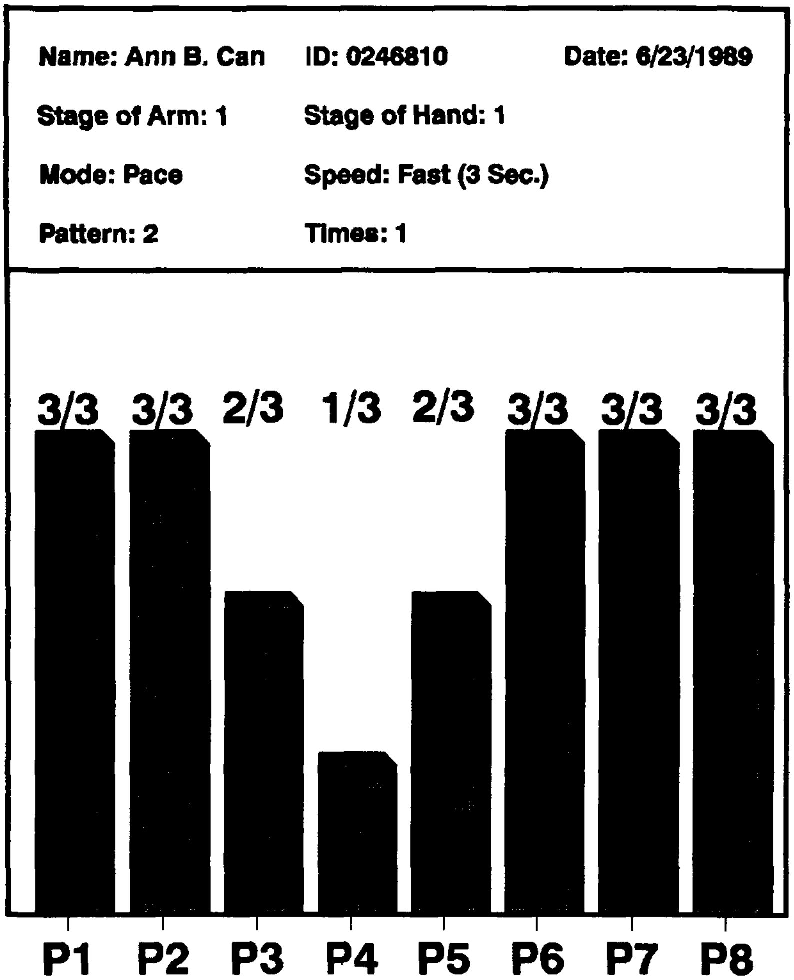

Figure 3.
Example of Pace Mode Report. P1 through P8 refer to the 8 points in space that make up the pattern. Each point is visited 3 times during each pattern run. The bar graph gives a visual report of the number of hits (0, 1, 2, or 3) for each point. The report also contains patient identifiers and information on the nature of the session: mode, speed, pattern, repetition of pattern. The (hypothetical) patient shown failed 2 out of 3 for point 4, and 1 out of 3 for points 3 and 5; the therapist may guess that the nature of the patient's motor problems underlies the lack of success for these points which are adjacent in space.

dynamic balance, attention, organizational skills, and ability to follow directions were suggested.

Some of these purposes would be better served by a modified or expanded robotic system. The staff made a number of specific suggestions in answer to a question (**Appendix B**, Q 27) on how they would like to see the system further developed (including a capability to program custom patterns, a module for using the gripper for fine prehension skills, and modules for cognitive training and auditory feedback).

Therapist acceptance

Seven therapists indicated that at first they were hesitant to use the robot aide due to a lack of familiarity with computers and robots, but that they felt more comfortable with time. Three never were hesitant to use the system. One therapist indicated that she never felt comfortable using the system because she could not adapt the program to obtain the patterns and speeds needed by her patient.

Ten of the therapists indicated that their *overall* response to the technology was positive. A list of the pros and cons they mentioned is contained in **Table 3.** It appears that most of the negative items on this list are due to the

Table 3.
Pros and cons of the robotic aide as reported by staff occupational therapists who used the system*

Pros	Frequency
Robot maintains patient's attention/is interesting (in doing repetitive exercises)	6
Robot provides good exercise/patients work harder and reach further	4
Appropriate for high and low stages of recovery; various patterns; easily graded	2
Provides (over-time, comparative) performance date	3
Provides an addition to treatment	1
New and interesting concept	1
Predictable and controllable patterns are provided	2
Patients feel it's good for their arm	1
Cons	**Frequency**
Robot exercises are boring	2
AROM only — no weights for resistance	1
Easy for the patient to cheat	1
Robot frightening/intimidating to patient	1
Cannot use system with low-level patients	1
Set-up takes too long (cuts into therapy time)	2
Proper use hard to remember	1
Not challenging for high level patients	1
Limited in present stage — may not be cost efficient	1
Not enough therapist control over pattern selection	1

*Based on response by 11 therapists

fact that the therapist wanted to use the system for patient problems for which it was not designed. This and other problems can be solved by further system development.

When asked if they would have one of these robot aide systems in their occupational therapy clinic if they were a director, one therapist answered with an unqualified "yes." All others specified that they would need to see proof of cost-effectiveness, or that the system ought to be further developed to include exercise programs for patients with other diagnoses in order to make it worth the price. Because a price and cost-effectiveness study is not yet available, the answers cannot be more than a rough indication of the perceived value of the system.

DISCUSSION

The purpose of our study was to determine whether the robot system was safe for patient and therapist, and acceptable to both of them. The data presented suggest that the system was considered safe. We did not experience a single safety incident, nor did therapists or patients express fears of accidents. Even though many of the patients were elderly, with limited education and cognitively compromised, they seemed intrigued by the system. Many expressed enjoyment working with it, and made other comments indicating that they accepted receiving treatment from a machine. Because this was an early test, the therapist was at the patient's side at all times; it may be that patients will feel less safe or interested if they work by themselves with the therapist nearby in the clinic area. Research on a larger number of patients is needed to explore this issue. Additional research is also needed to determine which patients can work with the system independently and for how long, thus allowing therapists to do other work.

While indicating their receptiveness to this innovative addition to occupational therapy practice, the therapists maintained a critical stance. Their reservations had their origin in two sets of factors: problems with the equipment (especially initially, when the research team was still finding and fixing bugs, and the therapists were not yet familiar with the system), and the shortcomings of the system (in its current version) as a therapy aide. To a degree, they forgot that the project was to provide proof-of-concept of

robot-assisted therapy, and immediately focused upon things the system could **not** do. The system that was tested was very basic; it did not offer the full range of movement patterns, nor full options for the therapists in order to select speeds, repetitions, and variations. We are working on new versions that will offer more choices for the therapist.

The therapists' critical stance is justified for other reasons. At this time there is no evidence to justify jumping headlong into therapeutic robotics. We do not yet know whether a robot can deliver therapy semi-independently, thus freeing the therapist to work in other areas. More importantly, we do not yet have information about the *outcomes* of robot-assisted therapy: is it as effective as traditional methods? Nor is detailed, objective time-trend documentation of patient performance at the current time perceived as an advantage by all therapists. This probably is due to the fact that occupational therapists have relied more on qualitative rather than quantitative data to assess their patients and monitor their progress.

A recent survey of 51 hospital-based occupational therapists without hands-on robotics experience found that they considered robots (in a PCA, workstation, etc., application) valuable, obedient, fun, and intelligent; however, they remained leery because robots are mysterious, difficult, and unfriendly (4). Our data suggest that occupational therapists can adjust quickly to working effectively with this type of sophisticated equipment.

The effectiveness and efficiency of equipment is a critical issue. We should not adopt new technology just because it is "high-tech," but because it is effective in treating patients or will save resources as compared to our current methods of treatment. Any equipment that is complicated, difficult to set up, quick to break down, or provides feedback that cannot be understood, is not likely to be in use for long, even though it may have been accepted initially because of the glamour of high technology.

Many therapeutic techniques provided by occupational and physical therapists consist of showing patients movements, guiding them through these, providing feedback on adequacy of performance, and then encouraging the patients to repeat the movement or task until it has been mastered or the therapeutic effect has been accomplished. (For many of these tasks, licensed therapists use assistants). These are things that a robot can do, relieving the therapists of tasks which can lead to boredom and burnout, and freeing up their time for more creative and satisfying aspects of their job. In principle, a robot system has many advantages when used in the routine aspects of therapy. It never tires, and is available 24 hours a day, 7 days a week. It never gets bored and will repeat the same activities with equanimity and without burnout. It has no negative emotional reactions towards the patients. It is never distracted and will maintain its level of concentration and diligence continuously.

These advantages are still to be realized, and most therapists will focus on the obvious disadvantages. Even the most advanced robot cannot offer solutions to unique patient or situational problems, and is unable to observe the patient's quality of movement, pain, refusal to cooperate, or other problems, and act accordingly. Thus, for patients with compromised competence (e.g., after stroke or traumatic brain injury), a therapist should be at hand. However, if the robotic system is designed in such a way that it cannot harm the patient, it appears possible for one therapist to monitor three or four robots, each administering therapy to one patient.

Usefulness and cost-effectiveness of a multiple-robot system will likely depend on the variety of applications available, so that the system can be used for the assessment and/or treatment of a number of patient problems. We already are using the hardware described above for two additional applications: "tracking," in which the patient is required to follow the robot end-effector and touch it within a specified time whenever the light goes on; and, "tapping," a test of the speed of tapping either one end-effector or alternating between the two. We have completed a prototype of a shape manipulating system that we expect to be useful for stroke patients who receive treatment for more advanced hand function. A project has just begun exploring the creation of modules for assessment and treatment of upper extremity coordination problems. These examples certainly do not exhaust the potential of treatment using robotics—modules for the assessment of hemineglect for example, or the treatment of problems involving strength or endurance, as well as many others are possible.

The more treatment applications are available using a robot system, the more there is a need for creative use of the system by the occupational therapist, who understands the patient's problem, as well as all treatment modalities and approaches. The robot will provide the repetition necessary for relearning and mastery; the therapist will assess, monitor, select modalities, and make decisions on how and how long to use them.

We consider the patients' response to the system positive (**Table 1**). It is surprising that so few called the treatment boring. One may doubt that patients will be inclined to do these exercises if they are administered as part of

routine treatment in rehabilitation, rather than in an experimental project for which the patient was invited and for which explicit written consent was given. It is certainly possible that the "Hawthorne effect" explains part of the patients' interest, acceptance, and willingness to work with the system. The interest in the system by the occupational therapy staff assigned to other patients, and other hospital staff and visitors was likely to stimulate the subjects in our study. However, in most cases the novelty wore off rather quickly, especially for those who had multiple sessions, and was replaced by the hard work of therapy. Research is needed to determine whether compliance is as good in routine use of the system as in an experiment.

Similarly, research is needed to demonstrate the efficiencies that can be attained with the system. A basic question is how much human attendance is necessary (e.g., can one therapist with one certified OT assistant "run" five or six systems?). Even if further field trials indicate that it is necessary for the therapist to work one-on-one with the patient, thus generating no cost savings, it may be that the ability of the robot aide to deliver standard movement patterns in a standard manner in the exact sequence and repetition seen as necessary, and generate objective, quantitative reports on the patient's performance, will enhance the occupational therapist's effectiveness enough so it is worth the price tag.

ACKNOWLEDGMENTS

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APPENDIX A
Robotic Evaluation: Patient Feedback

Primary therapist to fill out with patient:

Check one:
First experience ☐
Week of discharge ☐

1. Did you like this treatment?	Yes ____	No ____
2. Do you feel it was helpful?	Yes ____	No ____
3. Was this boring?	Yes ____	No ____
4. Did you feel safe?	Yes ____	No ____
5. Was if confusing to use?	Yes ____	No ____

Additional comments which would be helpful:

____
____
____
____

Patient name: ______________
R.I. # ______________
Age: ______________
Sex: ______________
Date: ______________

APPENDIX B:
ROBOTICS STUDY FOLLOW-UP QUESTIONNAIRE

1. How long did it take to set up your patients?
2. What took the longest in setting them up?
3. Was this time satisfactory to you?
4. Was setting up your patients complicated? If so, explain.
5. Were the markers on the floor helpful in set up? If not, explain.
6. Were the directions on the computer screen easy to follow? If not, explain.
7. What would you change on the screen to make it easier to follow?
8. Did the computer program work consistently upon your command? If not, explain problem.

9. Did the robot arm (not the computer) malfunction at any time? Please describe incidents.
10. Did you feel at any time that your patient was at risk of being hit by the robot arm? If so, what could be done to prevent this?
11. Would you feel comfortable leaving your patient alone to work? If not, what type of supervision would be needed?
12. Did you feel hesitant to use this system initially due to a lack of computer or robotic knowledge/experience? If yes, do you still feel the same since using it?
13. Did you ever use the data collected? Please explain your answer in detail.
14. Was the data confusing? If so, explain.
15. Would any other data have been more useful? Explain.
16. How often did you generate a printed copy of this data?
17. Was it easy to generate a report from the printer?
18. Overall, would you say the patients' responses to the robot were positive or negative?
19. Please list memorable patient responses both pro and con.
20. Overall, was your response to this positive or negative? Please list your pros and cons.
21. Were the available patterns adequate to meet your needs? If not, explain.
22. Were you able to obtain therapeutic movements from the established patterns?
23. Would you have preferred to program your own patterns?
24. Do you feel this was a valuable treatment modality for motor relearning? Why or why not is it valuable?
25. What do you see as its limitations at this point?
26. What type of patients would benefit from this modality in its current stage?
27. How would you like to see it further developed?
28. If you were an OT director would you want to order one of these systems for your clinic? Why or why not?

Department of Veterans Affairs

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Relative performance of single-channel and multichannel tactile aids for speech perception

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Abstract—Although the results from a number of studies of the performance of multichannel tactile aids for speech perception have suggested that such devices might provide more benefit to hearing-impaired persons than single-channel tactile aids (3,4), recent studies involving direct comparisons of multichannel and single-channel vibrotactile aids (5,6) indicated otherwise. In fact, for some types of speech information, such as rhythm and stress perception, single-channel aids were shown to be superior. The present study attempted to address this apparent discrepancy by comparing the performance of two single-channel devices with two multichannel devices in a variety of speech perception tasks including both single-item and connected speech stimuli. Results indicated that the two classes of tactile device performed similarly in rhythm and stress perception, but that the multichannel aids in many cases showed better performance for tasks in which the identification of fine-structure phoneme information was required (both single-item and connected speech). Results are discussed in terms of the possibility that the performance of a specific multichannel tactile aid cannot be considered indicative of all devices of the same class.

Key words: *artificial hearing, sensory aids-deaf, single-channel and multichannel devices, tactile perception, tactile vocoder.*

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INTRODUCTION

Devices that transform sound into tactile stimulation in order to convey sound and speech information to profoundly hearing-impaired persons are often classified by the number of tactile stimulators employed. Thus, single-channel tactile aids present information about the acoustic waveform via a single tactile transducer, and multichannel tactile aids utilize a number of tactile transducers. The design of many multichannel devices has been based on the assumption that the limited spectral resolution capability of the tactile system (14,23) would seriously impair the recognition of spectral characteristics of the vibratory signal of a single-channel device. Thus, in many multichannel tactile aids input acoustic frequency is recoded into location on the skin surface, substituting a dimension along which the tactile system shows good resolution ability (20,26). It was anticipated that such a strategy would permit some degree of recognition of spectral aspects of the acoustic signal, and thus lead to better performance than that found with single-channel devices. However, devices that have found their way into sustained commercial production have typically been of the single-channel or two-channel variety, due to the difficulties inherent in producing a wearable multichannel tactile aid (24). Studies investigating the relative effectiveness of single-channel and two-channel aids* have indicated that the addition of a second channel does not significantly increase subjects' ability to identify

*Broadstone SM, Weisenberger JM, Kozma-Spytek L. Relative benefits to speech perception provided by single-channel and two-channel tactile aids. Unpublished study. Central Institute for the Deaf, 1988.

phonemes or connected speech material, except in tasks specifically designed for the detection of fricatives.

The assumption of better performance with multichannel devices becomes important in light of recent clinical studies that compare performance of cochlear implants and tactile aids. In such studies, subjects are fitted with currently commercially available devices, and are tested to determine the benefits provided by these devices. The cochlear implant of choice in many of these recent studies has been the Nucleus 22-channel implant (Cochlear Corp). Morever, the lack of commercially available tactile aid options has necessitated the selection of a considerably less sophisticated device, the Tactaid II (Audiological Engineering) for tactile aid testing. In studies comparing the 22-channel implant with the 2-channel Tactaid II, the implant has consistently yielded dramatically higher levels of performance (13,20,22). The results of such comparisons might lead to the conclusion that tactile aids in general are not of significant benefit to hearing-impaired persons. Given the considerable discrepancy in the information available from 22-channel implants and 2-channel tactile aids, such a conclusion may be premature. Studies involving direct comparisons of single-channel or two-channel tactile devices with multichannel tactile devices are of some relevance in considering this issue.

Initial work comparing single-channel and multichannel tactile aids did not suggest that multichannel devices yielded any greater benefits than single-channel or two-channel instruments. In laboratory studies comparing the effectiveness of single-channel and multichannel tactile aids in conveying both phonemic and suprasegmental information from speech, Carney (5) and Carney and Beachler (6) evaluated the single-channel Fonator (Siemens Hearing Instruments) and a 24-channel vibrotactile device that provided a linear spectral array of stimulators (11). In phoneme recognition tasks, both devices yielded similar levels of performance under both tactile aid-alone and lipreading-plus-tactile aid conditions (it might be noted that for some stimuli, i.e., vowels, performance under the lipreading-alone condition was so high that any differences between devices in the lipreading-plus-tactile aid conditions may have been obscured by ceiling effects). In tasks involving the recognition of suprasegmental features such as syllable number, syllable stress, and intonation, the single-channel Fonator was found to be significantly better than the multichannel instrument. These results suggested that the assumption that a multichannel device would produce better performance was incorrect.

Lynch *et al.* (16) focused their attention on performance with connected speech. In their study, one profoundly hearing-impaired adult was tested in connected discourse tracking with two devices, the two-channel Tactaid II (Audiological Engineering, Inc.) and the 16-channel Tacticon TC-1600 (Tacticon Corp.), a 16-channel electrotactile linear display. Although the Tactaid II is technically a multichannel device, it is similar to a single-channel device in that one channel delivers primarily stimulus envelope information, while the second channel provides some indication of high-frequency activity, such as frication. Thus, while this study did not provide a direct comparison of a single-channel and a multichannel device, it might be anticipated that a 16-channel aid would provide considerably more information than a 2-channel aid. In Lynch's study, tracking performance with the Tacticon was *not* superior to that with the Tactaid II; in fact, the Tactaid II generally produced equal or higher tracking rates. Such a result suggests that the use of a multichannel tactile aid in a complex task such as connected discourse tracking does not provide more information than that available with a two-channel device.

The results are disturbing in view of other studies evaluating only single-channel or only multichannel tactile aids, the results of which suggest that multichannel devices *do* provide more information than is available with single-channel devices. For example, Brooks and her colleagues (2,3,4) evaluated the performance of a 16-channel linear vibratory device (the Queen's University tactile vocoder), and found high levels of phoneme and word recognition under tactile-aid-alone conditions. In addition, this device was also beneficial in recognition of connected speech in open-set sentences and connected discourse tracking, yielding an improvement in the tracking task of over 30 words per minute (wpm) in the lipreading-plus-tactile aid condition over lipreading alone. Similarly, Weisenberger *et al.* (27) found good phoneme recognition with the Queen's aid and the Tacticon TC-1600. Further, Weisenberger *et al.* (27) reported improvements in the connected discourse tracking task of 10 to 15 wpm for the Tacticon and 40 to 50 wpm for the Queen's aid, over lipreading alone. Regression analysis of the data from these two devices indicated a significant difference in the amount of benefit provided.

In contrast, evaluations of single-channel devices have generally shown less impressive results. For example, Weisenberger and Russell (28) found only fair recognition of vowels and poor recognition of consonants in limited sets under tactile-aid-alone conditions for subjects using two single-channel devices, the Siemens Minifonator and AB Special Instrument Minivib3. Miyamoto *et al.* (18)

and Skinner *et al.* (25) reported connected discourse tracking improvements of 5 wpm or less with single-channel devices, a considerably smaller degree of improvement than those listed for the multichannel devices above. In evaluating the ability of a single-channel device to convey intonation and contrastive stress, Bernstein *et al.* (1) found improvements over visual-alone conditions that were significant, but quite small (e.g., improvement in percent correct identification of intonation pattern of 3 percent under visual plus tactile aid conditions).

Further, Carney and Beachler's (6) finding that a single-channel device produced better performance in syllable number and stress tasks than that produced by a multichannel device was not corroborated by Kozma-Spytek and Weisenberger (15) in a case study of the multichannel Tacticon and single-channel Minivib3. In their syllable number and stress task, little difference between the two devices was seen, with slightly *better* performance in the Tacticon condition.

The present study was performed in an attempt to address further the question of the relative benefits available with single- and multichannel tactile aids. Both single-channel and multichannel tactile aids were evaluated, using both phoneme and connected discourse tasks, as was done by Carney (5). However, in contrast to the method used in the Carney studies, in which different groups of subjects were tested with each tactile aid, both single-channel and multichannel devices were evaluated in the same subjects, to minimize any intersubject differences that might affect performance.

METHOD

Subjects

A total of six undergraduate students (four men and two women) were paid for their participation. All had normal hearing as measured by audiometric test, and all subjects wore EAR foam earplugs and headphones through which pink masking noise was presented (80 dB SPL) to minimize acoustic cues. Because of the long-term nature of the testing, some subject attrition occurred; therefore, not all subjects participated in all tests.

Apparatus

Tactile aids. The first of the single-channel devices employed in the present study was the Minifonator, manufactured by Siemens Hearing Instruments, but no longer commercially available. The Minifonator has a wrist-worn circular electromechanical transducer with a contactor area of 2.5 cm^2. In the present experiments, the transducer was worn on the dorsal side of the left wrist in the area where a wrist watch is normally worn. The transducer is tethered by a cable to an 8.5 × 8 × 3.3 cm electronics package. The device is driven by four AA batteries. Controls accessible to the wearer include on/off, microphone gain, and vibratory intensity. Acoustic stimuli are detected by a small microphone, which is tethered to the electronics package via cable. Microphone gain and vibratory intensity were set to their highest settings, consistent with the observed preferences of naive tactile-aid users. The Minifonator processor provides a broad-band vibratory signal that preserves aspects of the spectral and temporal content of the acoustic signal.

The second single-channel tactile device used in the present study was the AB Special Instrument Minivib3. The Minivib transducer is rectangular in shape, with dimensions of 6.5 × 4.3 × 1.7 cm, and is tuned to resonate at a frequency at or near 250 Hz. In the present experiments, the Minivib was also worn by subjects on the dorsal side of the left wrist, using the wrist strap included with the device for such purpose. The transducer is connected by a cable to a 1.5 × 2.0 × 1.0 cm electronics package, with a built-in miniature microphone. The device is powered by a specially constructed rechargeable battery whose output is approximately 10 V. Controls available to the user include on/off and microphone gain; vibratory intensity and transducer tuning can be modified by turning a screwdriver in two recessed pots located on the bottom of the device. The gain of the Minivib is quite strong, in that setting it to its maximum of nine permits the detection of sounds occurring at considerable distance. For this reason, the gain of the Minivib was set to one-half its range (4.5) for the present experiments. Vibratory intensity was left at the factory setting. The processor of the Minivib performs an envelope extraction on the incoming acoustic waveform and uses this envelope to modulate the amplitude of a 250-Hz carrier, which drives the vibratory transducer. Thus, this device is a relatively straightforward AM processor.

The first of the multichannel tactile devices evaluated in the present study was a microprocessor-based implementation of the Queen's University vibrotactile vocoder developed at the Central Institute for the Deaf (10). A description of the original device can be found in Scilley.*

*Scilley PL. Evaluation of an auditory prosthetic device for the profoundly deaf. Unpublished master's thesis. Queen's University, Kingston, Ontario, 1980.

In the CID implementation of the vocoder, input acoustic stimuli are detected by an ACS headset electret microphone and then passed through a logarithmic gain function and high-frequency preemphasis circuit to a bank of 16, one-third octave switched-capacitor filters, with center frequencies between 140 and 6350 Hz (the two lowest-frequency channels have a bandwidth of two-thirds of an octave). The envelope output of each of these filters is used to modulate the amplitude of a 100-Hz square wave, which drives one of 16 magnetic solenoids mounted in a 19.5 cm linear array worn on the underside of the forearm. The center-to-center distance between solenoids is 10 mm. The output voltage to the solenoids is proportional to the input to each channel; overall level is determined internally and is not variable by the subject. Perceived stimulation levels vary over a range of approximately 0 to 40 dB SL, with an average perceived level of approximately 20 dB SL. As worn by the subjects, low-frequency stimulation is felt near the wrist, and high-frequency stimulation is felt near the elbow.

The second multichannel tactile device evaluated was the Tacticon Corporation TC-1600, an electrotactile vocoder that was formerly commercially available, but which has been discontinued. This device is functionally similar in most major respects to the Queen's vocoder, although there are several important differences. Input acoustic stimuli are detected by a Sony ECM 16-T electret microphone and passed through a bank of 16 logarithmically spaced filters, with center frequencies between 100 and 7,000 Hz. The envelope output of each filter modulates the pulse rate of a biphasic current pulse, each phase having a duration of 20 ms and nominal current of 7 to 12 mA as set by the user's display intensity control. Pulse rates vary between approximately 0 and 500 pps. This pulse is transmitted to one of 16 electrodes, arranged in a 22.5 cm array worn on the abdomen. The center-to-center distance between electrodes is 15 mm. A noise-suppression circuit in the processor was operated with an attack time of 5 seconds, and recovery time of 5 to 10 seconds. The primary effect of this circuit is on sounds in the 1.5 to 4 kHz range. The processor and battery enclosure for the device measures 15.5 × 9 × 4.5 cm and weighs approximately 680 g. The processor box can be worn clipped to a belt, in a backpack, or elsewhere, and in the present study was placed on a table next to the subject. The level of stimulation on the Tacticon is controllable by the subject with a dial on the processor box and was set by the subject to a comfortable level at the beginning of each test session. The surface of the abdomen was moistened with water prior to putting on the device.

Stimuli. All stimuli were presented live-voice by three female talkers. All subjects had some exposure to at least two of the three talkers, although the number of sessions with each talker was not equal for all subjects.

Several tasks were employed, arranged in a hierarchy from simplest to most difficult. The first task was minimal pairs phoneme discrimination. Four sets of stimuli were employed. All items in each set consisted of a pair of words that differed in only one phoneme. In the first set (40 items), the initial phoneme of each pair differed in manner of articulation, with place articulatory features (and voicing, where possible) held constant (e.g., tea-see). In the second set (40 items), the initial phoneme differed in place of articulation, with manner and voicing features held constant (e.g., bait-gate). In the third set (20 items), the initial phoneme differed in consonant voicing, with place and manner features held constant (e.g., bat-pat). In the fourth set (8 items), items in a pair differed in their medial vowel, in an /hVd/ format. Stimuli were presented in isolation (i.e., no carrier phrase was used).

The second task was syllable rhythm and stress identification, tested in a manner similar to Erber's monosyllable-trochee-spondee test (12). Fifteen items were used, including five monosyllables, five trochees, and five spondees. Subject responses were scored twice: once to determine whether the correct item had been selected (identification), and again to determine whether a word with the correct syllable number and stress had been selected (categorization).

The third task was integration with lipreading. Sets of eight items differing in either their initial consonant or their final consonant were presented. The initial consonant lists were presented in a /Cat/format, and the final consonant lists were presented in an /aC/ format. Lists were tested both in isolation and in the context of fixed carrier phrase, "Tell me __________, please."

The final task was connected discourse tracking (9). The text material used was *Star Trek IV: The Voyage Home* (17), a low-difficulty adult text, presented in 5-minute tracking sessions.

Procedure

Because it was felt that exposure to all four devices would not permit sufficient training time with any one device, all subjects were tested with two devices, one single-channel and one multichannel. The assignment of single-channel devices and multichannel devices to subjects was random. During a test session, the subject wore one device

for half of the testing time and the other device for the other half of the testing time. The order of test condition was varied across test sessions.

Minimal pairs. The first five sessions were considered training sessions. On each trial in a training session an ABX design was used, in which the talker read each of the words in the pair and then presented one of the words. The subject indicated which item (A or B) had been presented by saying "A" or "B," followed by the chosen word. The experimenter provided feedback by presenting the correct word acoustically through the intercom system. Each item in a list was presented in the above fashion, and each list was presented four times under each testing condition in the course of a session. Subject responses were recorded on a score sheet.

The last five sessions were considered test sessions. On each trial an ABX design was not used and only the test word was presented. Subject responses and feedback were provided as above. As in the training sessions, each list was presented four times.

Syllable rhythm and stress. The first 10 sessions were considered training sessions. On each trial, the talker presented a word selected at random from the 15-item test list, and the subject responded with the word that had been presented. Feedback was presented acoustically. Each item in the list was presented five times in the course of a session, under each testing condition. The last five sessions were considered testing sessions, and were run in an identical manner.

Integration. For stimuli presented in isolation, five conditions were tested: single-channel aid alone, multichannel aid alone, lipreading alone, lipreading plus single-channel aid, and lipreading plus multichannel aid. For stimuli presented in carrier phrases, the two tactile-aid-alone conditions were omitted. Under all conditions involving lipreading, the shade on the window of the booth was raised. Subjects were first tested with stimuli presented in isolation. Following completion of this testing, they were then tested with stimuli in carrier phrases. For each of these phases of testing, the first five sessions were considered training sessions, and the last five were considered test sessions. Within each session, each list (initial and final consonants) was employed. On a trial, the talker presented a word from the list, and the subject responded with a word from the list. Feedback was delivered acoustically. For each testing condition, each stimulus was presented five times during a session.

Tracking. The connected discourse tracking procedure outlined by DeFilippo and Scott (9) was used. In this procedure, a talker reads aloud a portion of meaningful text (word, phrase, or sentence) at a normal speaking rate, and a receiver attempts to repeat the phrase verbatim. If the receiver does not repeat the phrase correctly, the talker may use strategies such as repeating some or all of the phrase, reviewing a previous phrase or sentence, or providing context clues by reading the next phrase or sentence. Specific examples of each strategy are provided in DeFilippo and Scott. If the response is correct, the reader presents the next phrase. In the present experiment, 5-minute tracking sessions were used. The total number of words correctly repeated by the subject during each 5-minute session was counted and divided by the total number of minutes in the session, to yield a word-per-minute (wpm) score.

Three conditions were tested: lipreading alone, lipreading plus single-channel aid, and lipreading plus multichannel aid. Each subject was tested daily in a 1-hour test session. During each test period, five tracking sessions were obtained: one under the lipreading-alone condition, and two each under the lipreading-plus-single-channel aid and lipreading-plus-multichannel aid conditions. The order of testing was varied across days to eliminate order effects. For this testing, a single female talker presented text. A total of 60 5-minute tracking sessions was obtained over a period of approximately 4 weeks.

RESULTS AND DISCUSSION

Minimal pairs

For clarity of presentation, overall results for all tasks are averaged across single-channel devices and across multichannel devices. No significant differences were obtained between single-channel devices or between multichannel devices for the tasks described above. Results for five subjects in the minimal pairs task are shown in **Figure 1**. A two-way, within-subjects analysis of variance was performed on arcsine-transformed percentage scores to determine the significance of differences in performance. The results of this analysis indicated that there was a significant overall effect of type of tactile device (single-channel versus multichannel) ($F(1,4)=8.78$, $p<0.05$), and significant differences among type of stimulus set (manner, place, voicing, and vowels) ($F(3,12)=12.24$, $p<0.001$). Post-hoc analysis by Tukey's test showed that manner features were perceived significantly better than place (0.25, $p<0.01$), voicing (0.19, $p<0.01$), and vowels (0.25, $p<0.01$). No other comparisons were significant. In addition, no inter-

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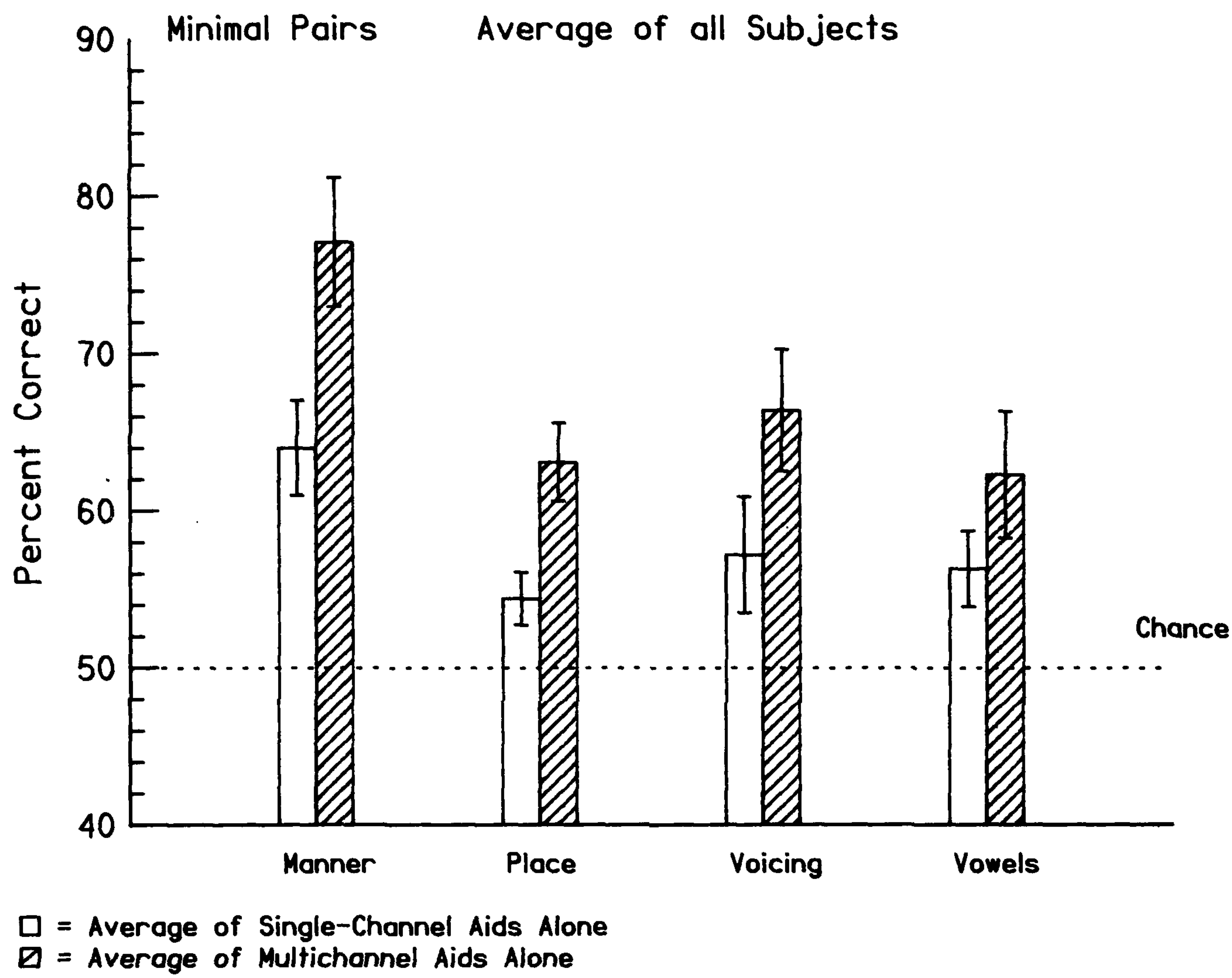

Figure 1.
Averaged percent correct performance of five subjects with single-channel and multichannel tactile aids for the last five sessions of the minimal pairs task, as a function of type of articulatory contrast.

action effect was observed in the analysis of variance ($F(3,12)=0.37$, ns).

These analyses indicate that subjects received significantly more information in this task from the multichannel devices than from the single-channel devices. The lack of an interaction suggests that this improvement was present for all classes of stimuli. The finding that manner features were transmitted more effectively than the other types of stimuli is consistent with previous findings for multichannel devices (3,27).

Syllable rhythm and stress

Results for five subjects are shown in **Figure 2** for identification (selection of the correct word) and categorization (selection of the correct syllable number an stress). As can be seen, performance was similar with both types of device, and no significant differences were observed with a dependent-groups t-test on arcsine-transformed percentages for either identification ($t(4)=0.55$, ns) or categorization ($t(4)=0.24$, ns). Carney and Beachler (6) found that a single-channel device yielded better performance in such suprasegmental tasks than a multichannel device. However, Kozma-Spytek and Weisenberger (5) found in a case study with one hearing-impaired child that, while the Tacticon multichannel aid was comparable to the single-channel Minivib in categorization, substantially better identifica-

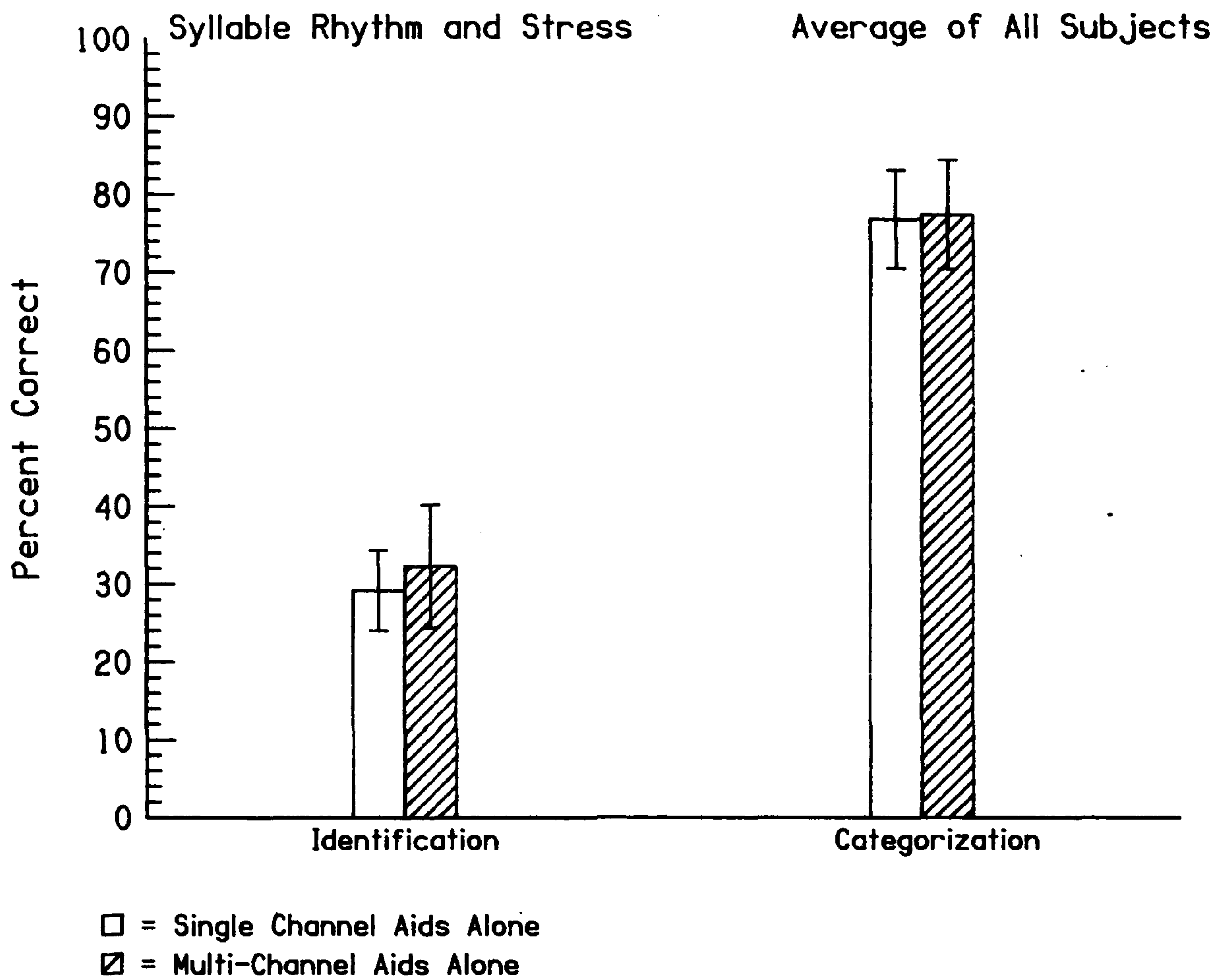

Figure 2.
Averaged percent correct performance for five subjects with single-channel and multichannel tactile aids on the syllable rhythm and stress task, for stimulus identification and rhythm and stress categorization.

tion performance was observed for the Tacticon than the Minivib. It might be noted that the present results are not consistent with either of these previous findings.

Integration

The results for the integration task, completed for five subjects for stimuli in isolation and for three subjects for stimuli in carrier phrases, were somewhat more complex. Data for stimuli presented in isolation and in carrier phrases will be discussed separately. For initial consonant stimuli presented in isolation (**Figure 3a**), a within-subjects analysis of variance for arcsine-transformed percentages yielded a significant overall effect of testing condition ($F(4,16)=28.43$, $p<0.001$). Post-hoc analysis by Tukey's test showed that the single-channel and multichannel aid alone conditions were significantly lower than lipreading alone (single-channel: 0.49, $p<0.01$; multichannel: 0.31, $p<0.05$), and each device-alone condition was significantly lower than the corresponding lipreading-plus-device condition (single-channel: 0.57, $p<0.01$; multichannel: 0.71, $p<0.01$). However, no significant difference between the single-channel and multichannel aid alone conditions was found.

In assessing the benefits of each type of device to lipreading, it was found that the lipreading-plus-single-channel condition was not significantly different from

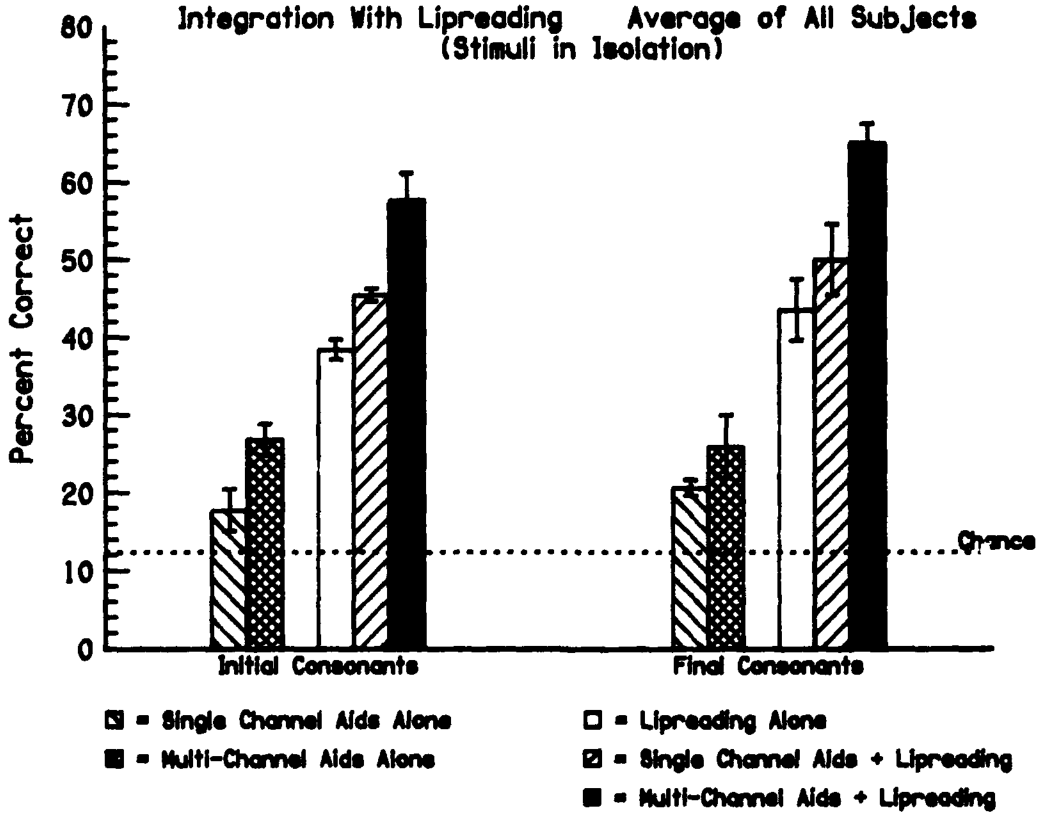

Figure 3A.
Averaged percent correct performance for five subjects with single-channel and multichannel tactile aids on the integration task. Data for identification of initial-consonant and final-consonant stimuli. Results for stimuli presented in isolation.

lipreading alone, whereas the lipreading-plus-multichannel condition showed a significant improvement over lipreading alone (0.40, $p<0.01$). Further, the lipreading-plus-multichannel condition was significantly higher than the lipreading-plus-single-channel condition (0.32, $p<0.05$).

A similar pattern of results was obtained for the final consonants presented in isolation ($F(4,16)=19.28$, $p<0.001$ overall effect of testing condition). Post-hoc testing revealed similar results to those for the initial consonants, with the exception that the lipreading-plus-multichannel condition was *not* significantly different from the lipreading-plus-single-channel condition. Nonetheless, it was still the case that a significant difference was obtained between lipreading alone and lipreading-plus-multichannel conditions (0.27, $p<0.05$), and not between lipreading alone and lipreading-plus-single-channel conditions.

For stimuli presented in carrier phrases (**Figure 3b**), analysis of variance for the initial consonants showed a significant effect of test condition ($F(2,4)=10.24$, $p<0.05$). Post-hoc testing showed only one significant comparison, that between lipreading alone and lipreading-plus-multichannel conditions (0.35, $p<0.05$).

For the final consonants presented in carrier phrases, no significant effect of test condition was observed ($F(2,4)=5.01$, ns). Post-hoc testing was not performed. Although it can be seen in **Figure 3b** that the lipreading-plus-multichannel condition produced slightly higher levels of performance, the differences across conditions are not large.

Overall, the results for the integration task suggest that the multichannel devices provided more assistance to lipreading than did the single-channel devices. Interestingly, there were in many cases no significant differences between performance in the two device-alone conditions, suggesting that performance in the lipreading-plus-device conditions is not a simple addition of device alone and lipreading alone performance.

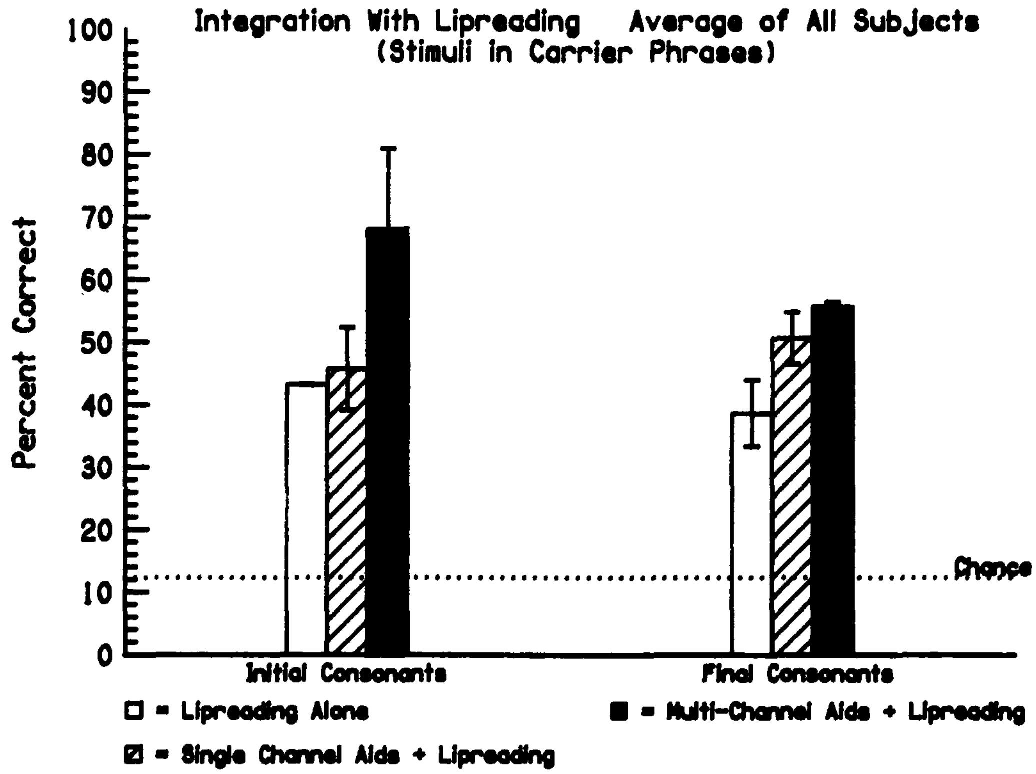

Figure 3B.
Averaged percent correct performance for five subjects with single-channel and multichannel tactile aids on the integration task. Data for identification of initial-consonant and final-consonant stimuli. Results for stimuli in carrier phrases.

Tracking

As a result of subject attrition over the long duration of testing employed in the present study (daily testing for several months), only one subject was available for testing in the connected discourse tracking task. This subject was tested with the single-channel Minivib and the multichannel Queen's vocoder. Results for this subject are shown in **Figure 4** for lipreading alone, lipreading-plus-single-channel, and lipreading-plus-multichannel conditions. As can be seen, the single-channel aid produced a slight improvement in performance over lipreading alone, on the order of about 5 wpm. Performance was much higher with the multichannel device, with an improvement over lipreading alone of closer to 20 wpm.

To test whether these performance differences were statistically significant, regression lines were fitted to the data for each condition using a least-squares technique. The slopes of the regression lines were compared in pairwise *t*-tests (7). These comparisons indicated significant differences between the lipreading-plus-multichannel condition and lipreading alone ($t(26)=2.089$, $p<0.05$), but no differences between the lipreading-plus-single-channel and lipreading alone conditions ($t(26)=0.49$, ns).

Although it is speculative to draw conclusions from the data for a single subject, it might be noted that performance under the lipreading-plus-multichannel condition is in agreement with data reported by Brooks *et al.* (4) and by Weisenberger *et al.* (27) on the degree of improvement provided by multichannel devices in connected discourse tracking. Further, the results for this subject in the lipreading-plus-single-channel condition are in good agreement with data reported by researchers such as Miyamoto *et al.* (18).

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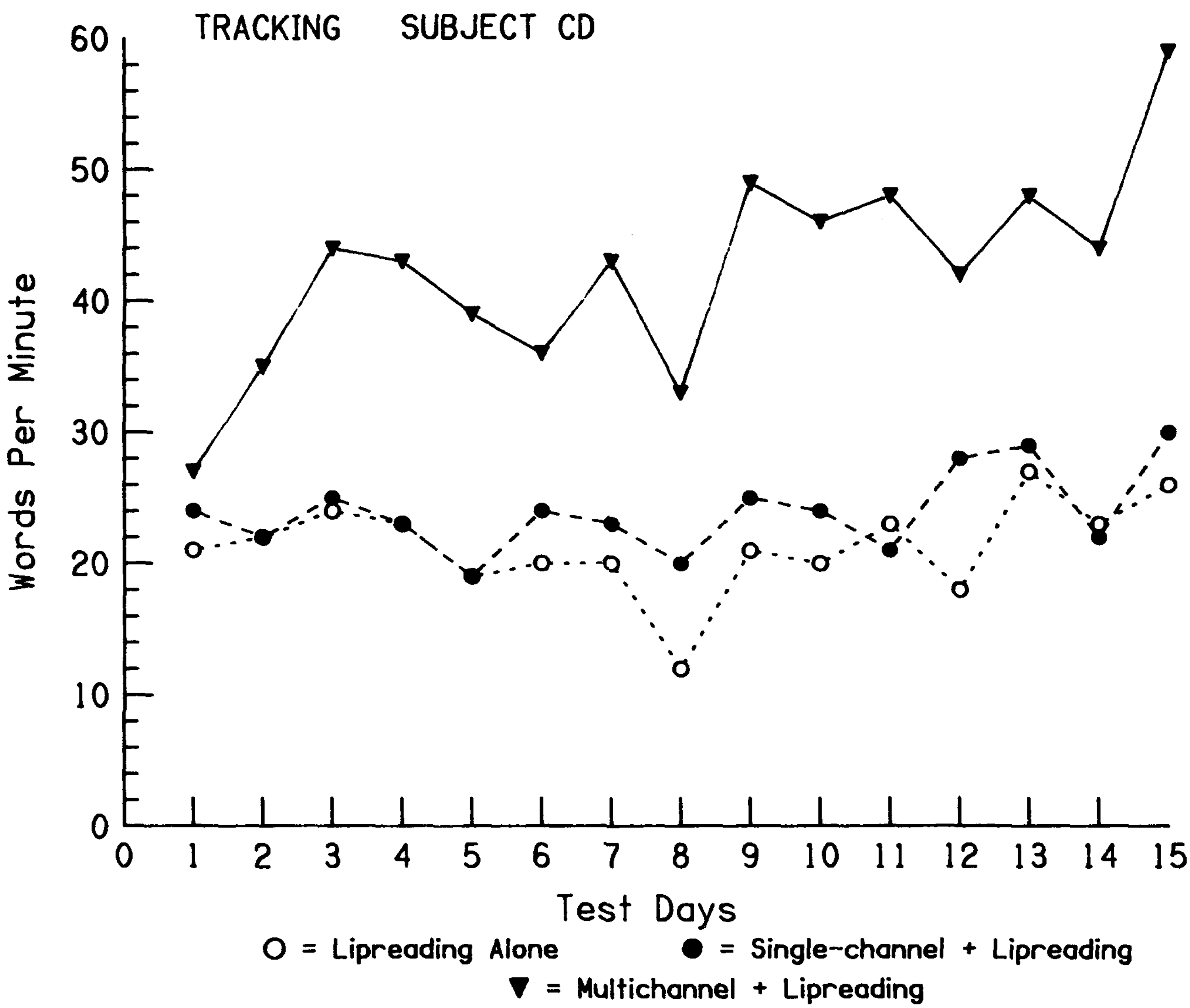

Figure 4.
Tracking performance for one subject with single-channel and multichannel tactile aids, over test days.

GENERAL DISCUSSION AND CONCLUSIONS

Overall, the results obtained in the present study suggest that more information is provided in many cases by multichannel tactile aids than by single-channel aids in speech perception tasks requiring phoneme recognition, but that the two classes of device perform comparably in tasks requiring the perception of amplitude envelop characteristics, such as the syllable rhythm and stress task.

In general, the present results are not consistent with the findings of Carney and Beachler (6) and Carney (5). However, there are a number of differences between those studies and the present study that may have contributed to differences in the obtained results. For example, slightly different tasks were employed. In addition, the present study utilized a within-subjects rather than independent groups experimental design, and a longer duration of training was used in the present study. Perhaps more importantly, the two studies employed different tactile devices. Support for the suggestion that device differences may be important may be found in studies comparing several single-channel or several multichannel devices. For example,

Weisenberger *et al.* (27) compared two 16-channel tactile aids, the Queen's vocoder and the Tacticon TC-1600, in a series of phoneme recognition and connected speech tasks. While the two devices performed comparably on the phoneme recognition tasks, the Queen's vocoder showed much higher performance in tracking, yielding improvements over lipreading alone of more than 40 wpm, as compared to the approximately 10 wpm provided by the Tacticon. Similarly, the tracking data reported by Lynch *et al.* (16) for the Tacticon show rather poor performance for the Tacticon, and further data collected in our laboratory support this finding. In addition, Rakowski *et al.* (21) found poor tracking performance for the Audiotact (Sevrain-Tech), a 32-channel electrotactile aid.

Such results suggest that not all multichannel tactile devices provide equivalent amounts of speech information to the wearer. It is possible that the 24-channel vocoder used by Carney and Beachler (6) and Carney (5) did not provide as much information as was provided in the present study by the Queen's vocoder and Tacticon. Although the two multichannel devices used in the present study did not show significant differences in the single-item tasks employed, it is important to note that the subject for whom tracking data are reported in the present study had been trained with the Queen's vocoder as the multichannel device. Had he been trained instead with the Tacticon, the difference in tracking results between lipreading-plus-single-channel and lipreading-plus-multichannel conditions may have been less dramatic. Conversely, it is also possible that the Fonator, the single-channel device used by Carney, is a superior single-channel aid that provides more information about speech than the Minifonator and Minivib used in the present study. Although the Minivib and Minifonator have also been found to be similar to each other in performance (27), and were not found to be different from each other in the present study, better performance may be attainable with the Fonator than for either of these two devices.

Perhaps the appropriate conclusion from consideration of the results of the present study and of the previous work is that some multichannel tactile devices can provide significantly more benefit to the wearer than some single-channel tactile devices. Generalizing the results obtained in a particular study to all devices of a particular class may or may not provide an accurate view of potential tactile aid benefit.

The data from the present study do suggest that results from comparisons of multichannel cochlear implants and two-channel tactile aids should be considered in view of the possibility that a multichannel tactile aid might provide greater benefits than are seen with two-channel device. As multichannel tactile aids become commercially available, this possibility can be tested empirically. Indeed, Osberger (19) and Robbins (22) reported promising pilot results for clinical trials of the Tactaid VII (Audiological Engineering), a seven-channel prototype tactile device, which was introduced into their cochlear implant testing program. In addition, Cowan *et al.* (8), in studies of the Tickler Talker, an eight-channel electrotactile aid, show comparable performance on some speech perception tasks with this device and a 22-channel implant. Although considerable further work of this nature is necessary before substantive conclusions can be drawn regarding the comparability of implants and tactile aids, the results of the present study, taken together with other reports of multichannel tactile aid performance, suggest that multichannel tactile aids may prove a viable option for speech perception by hearing-impaired persons.

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A Technical Note

Audible pedestrian traffic signals: Part 1. Prevalence and impact

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Abstract—A collaborative project between the San Diego Association of Governments and San Diego State University (5) evaluated the effectiveness of audible pedestrian traffic signals in aiding visually disabled and elderly persons to walk in their community with greater safety. Three aspects of audible pedestrian traffic signals were investigated: 1) the patterns of use and the impact of these signals on pedestrian traffic safety; 2) the physical characteristics of the sound emitted by the devices; and, 3) the detection of the emitted sounds in the presence of various traffic noise levels. This paper reports on the prevalence and impact of audible traffic signals were ascertained through seeking information from traffic engineers in 71 North American cities; soliciting opinions about these signals from various school officials, social agencies, and volunteer organizations that serve persons with vision impairments; and analyzing pedestrian accident rates at intersections before and after the installation of such signals. The other two aspects of the project are reported in accompanying articles (6,7) that appear in this issue of the *Journal of Rehabilitation Research and Development.*

Key words: *accident rates, audible pedestrian traffic signals, blind and elderly pedestrians, mobility and orientation instructors, reactions, surveys.*

INTRODUCTION

The objective of the project was to learn if audible pedestrian traffic signals helped visually disabled persons to cross streets more safely. To do this, a study was made of: 1) the prevalence and patterns of use and the impact of these signals on pedestrian travel; 2) the physical characteristics of the sound emitted by the devices; and, 3) the detection of the emitted sounds by various persons in the presence of various traffic noise levels. This paper will focus on the prevalence of audible pedestrian traffic signals in North America, their impact on pedestrian traffic safety in nine California cities, and the public's reactions to such signals.

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Description of audible pedestrian traffic signals

Various types of audible pedestrian traffic signals (APTS) have been used for a number of years in locales throughout the world, including the United States (2). During the past 10 years, their use in the U.S. has greatly increased due to the availability of several reliable and relatively inexpensive (under $300) devices. APTS are attached to vehicular traffic control signals and emit a distinct sound during the WALK phase of the signal (**Figure 1**). The sounds from the audible signals are intended to alert pedestrians to the WALK phase in the signal cycle thus encouraging them to begin promptly crossing the street before the light changes.

To use an audible signal, the pedestrian pushes the button to activate the WALK signal at the signalized intersection. During the WALK phase of the traffic cycle, the APTS emits one characteristic sound for the north-south direction and a different sound for the east-west direction. No sound is emitted during the flashing or steady DON'T WALK phases. At some intersections, the WALK sign is always active so that audible signals installed at these locations would sound for every traffic cycle. To avoid disturb-

ing nearby residents, timers can be added to turn off the audible signals at night.

Two brands of audible traffic signals that are commonly used in the United States and Canada are the Sonalert buzzers manufactured by Mallory Capacitor Company's Emhart Electrical/Electronic Group, Indianapolis, IN, and the audio pedestrian signal manufactured by Traconex, Inc., Santa Clara, CA, under license from Nagoya Electric Works of Nagoya, Japan. Both devices attach to the pedestrian signal box of the traffic signal and are wired to emit a distinct sound with the WALK or green portion of the signal.

The Sonalert is a small (4.3 cm diameter, 5.1 cm long cylinder) piezoelectric signaling device intended for installation *inside* the chassis of a crosswalk light box.* Installation of the Sonalert requires that a 3 cm hole be drilled into the crosswalk light box and the unit installed through this opening. According to its manufacturer (3), Sonalert Model SC616WY produces three different sound patterns: pulsating tones at 1750 Hz or 3000 Hz and a warble tone that switches between these two frequencies. Sonalert Model SC616WXY produces five different sound patterns: pulsating tones at 1750 Hz and 3000 Hz, continuous tones at these two frequencies, and a warble tone that switches between these frequencies. Both Sonalert units are controlled and powered from a 6-16 VDC source and can generate a sound pressure of 75 dB at 0.6 m (2 ft).

The Nagoya/Traconex APTS (8) is a self-contained, bell-shaped and weatherproof unit that attaches externally to the pedestrian crosswalk signal box (**Figure 1**). The Nagoya/Traconex signals emit an electronic "cuckoo" sound to denote the north-south direction of travel and an electronic "chirp" sound to denote the east-west direction of travel. The manufacturer lists the following specifications for the "cuckoo" sound: frequency base of 1100 Hz; sound duration of 0.6 seconds; and a frequency deviation (warble) of 120 Hz. The manufacturer's specifications for the "chirp" sound are a continuous frequency variation of 800 Hz that lasts 0.2 s with a frequency base of 2800 Hz. The output for both units is specified to be 90 dB sound pressure level (SPL, re: 20 μPa) at 1 m and self-switching between one of two adjustable output volume levels depending on the ambient noise level. Because the Traconex signal is so widely used in the western United States, especially in California, and because of numerous cooperative study sites in California, this research effort focused on the Nagoya/Traconex unit.

*Personal communication with Windsor P. Waits, Applications Engineer, Mallory Capacitor Company, Indianapolis, IN, June 22, 1989.

Beneficiaries of audible pedestrian traffic signals

The stated purpose for using audible traffic signals is to help visually impaired pedestrians travel more safely and confidently (8,11). Visually impaired and elderly persons may walk slowly and thus would need the full walk cycle to traverse the street completely before the signal changes. Therefore, it is beneficial for them to know definitely that they have the WALK signal so they can begin crossing the street without delay.

Visually impaired and elderly persons comprise a sizable population in the United States. The U.S. Bureau of the Census (9) reports that 12.8 million persons (or 7 percent of the U.S. population over 15 years of age) have visual disabilities, and 1.7 million (or 13 percent) of these have severe visual disabilities. Almost half (45 percent) of the population with vision disabilities are elderly, and 71 percent of those with severe visual dysfunctions are elderly. The number of persons over 65 has almost doubled in the last quarter century, and actuarial trends (9) show that the elderly population will grow even faster in the next 25 years.

A literature search yielded little on the prevalence and impact of audible pedestrian traffic signals. Hulshcer (1) examined problems related to providing an audible device for blind pedestrians. In Wilson's study (11), which was cited in an Issue Paper by the Institute of Transportation Engineers (2), pedestrians were found to cross the street more quickly when APTS were present. In his feasibility study of audible pedestrian signals, Oliver (4) reported that 103 U.S. cities and 34 foreign locations have some type of audible signals. Uslan (10) reported that his blind subjects reacted positively to audible signals, especially at intersections that had complex traffic flow patterns.

To ascertain the prevalence and impact of APTS, this project asked three questions: 1) How prevalent are audible traffic signals in North America? 2) Have these signals had an impact on pedestrian safety? and, 3) What are some of the reactions to these signals?

Answers were sought by contacting persons responsible for the installation of audible traffic signals, analyzing pedestrian traffic accident rates at intersections before and after the installation of audible traffic signals, and interviewing organizations that work with persons who have visual impairments.

PREVALENCE OF AUDIBLE TRAFFIC SIGNALS

To determine the numbers and locations of APTS, a simple fact-finding survey was mailed to traffic engineers

Figure 1.
The bell-shaped Nagaya/Traconex audible pedestrian traffic signal is mounted atop the pedestrian signal box and pointed downward.

in 71 North American cities reported to have one or more audible traffic signals in their jurisdiction. The survey sought information about their signal installations. Follow-up was made by telephone.

Of the 71 cities surveyed in 1988, 52 cities and various locations in Connecticut had audible pedestrian traffic signals. In the 52 cities and the State of Connecticut, there were 288 intersections with audible signals. Of these, 184 (64 percent) intersections have the Nagoya/Traconex units, with the great majority being in the western U.S. and Canada. In contrast, the unit by Mallory (or some similar buzzer system) tended to be more popular in the eastern United States.

The distribution of audible signals was quite uneven. Among the 52 cities identified as having APTS, 25 (47 percent) were in California. Nearly half the 52 cities had only one or two intersections with audible signals. Only seven cities, mostly in California, had audible signals at more than 10 locations. California cities had 129 intersections with audible signals. At the time of the survey, 29 of the 52 cities reported that they expected to install additional audible signals in the near future.

The survey respondents were also asked to identify the locations of their audible signals. The most common location for APTS was in business districts where 60 percent of the cities have installed them. The next most popular locations, with 38 percent each, were entrances to colleges or universities and intersections near service centers for persons with disabilities. Audible signals were also installed near schools (21 percent); residential areas (15 percent); medical centers (13 percent); government offices and seniors' service centers (6 percent each); and tourist attractions and transit centers (2 percent each).

Among the 52 cities with APTS, 43 (81 percent) placed at least one of their signals at intersections with pedestrian actuated controls (push buttons). Ten cities (19 percent) have timers attached to the signals to deactivate the signals at night.

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IMPACT ON PEDESTRIAN TRAFFIC SAFETY

Nine California cities were surveyed to learn about pedestrian accidents at the intersections with audible signals Pedestrian accident records were reviewed for the calendar year in which the signals were installed and for the two previous calendar years and the two calendar years after their installation. The pedestrian accident data from these nine cities are summarized in **Table 1.**

The nine cities were chosen on the following basis: 1) willingness of the city staff to participate in the research; 2) the availability of detailed accident records; 3) a minimum of 2 years of experience with audible signals; and, 4) the usage of the same equipment, the Traconex audible pedestrian signal. By limiting the choice of cities to one state and the type of APTS to the Traconex brand, the number of variables that could affect the data (e.g., the motor vehicle code, installation procedures for APTS, accident reporting requirements, and weather pattern) was minimized.

Table 1 shows that the nine California cities had a total of 58 intersections and two mid-block crossings with audi-

Table 1.
Impact of audible signals on pedestrian accident rates

City	Intersections				Number of		
	T	W	W-O	ACC	ACC	INJ	NO INJ
Beverly Hills	2	0	2		0	0	0
Cupertino	5	3	2		5	4	1
Huntington Beach	13	5	8		9	9	1
Los Angeles	4	2	2		4	6	0
Norwalk	3	1	2		3	3	1
San Diego	6	4	2		8	8	0
San Jose	1	0	1		0	0	0
Santa Monica	5	3	2		3	4	0
West Covina	21	9	12		16	13	3
TOTAL	60	27	33		48	47	6
Before installation of audible signals							
Cupertino					4	3	1
Huntington Beach					5	5	1
Los Angeles					2	2	0
Norwalk					2	1	1
San Diego					5	5	0
Santa Monica					0	0	0
West Covina					6	4	2
Total					24	20	5
After installation of audible signals							
Cupertino					1	1	0
Huntington Beach					4	4	0
Los Angeles					2	4	0
Norwalk					1	2	0
San Diego					3	3	0
Santa Monica					3	4	0
West Covina					10	9	1
Total					24	27	1

T = Total intersections
W = Intersections with accidents
W-O = Intersections without accidents
ACC = Accidents
INJ = Injured pedestrians

ble pedestrian signals. Of these, 31 intersections and the two mid-block crossings had no pedestrian accidents in the 5 years studied. Of the 27 (45 percent) intersections that had accidents, exactly half the 48 accidents occurred before the installation of the audible pedestrian signals and half afterwards. Twenty persons suffered injuries in the 24 accidents before the installation of the audible signals, including one multiple-injury accident. Twenty-seven persons suffered injuries after APTS installation, including three multiple-injury accidents. There were no injuries in six of the 48 accidents.

Of the 47 pedestrians injured in the 48 accidents, two accidents involved persons carrying a white cane. One of these accidents occurred before the installation of the audible signals and one afterwards. Both accidents occurred at intersections near service centers for disabled persons, and the driver/vehicle was at fault. There were no pedestrian deaths, and alcohol was not involved in any of the 48 accidents studied.

Age and sex of the injured pedestrians were detailed in 26 (54 percent) of the accidents. Ten injured pedestrians were in the 0-17 age group, with eight being injured in accidents before installation of the signal and two afterwards. In the over 18 age group, more females than males were injured before the installation of APTS. In the accidents which occurred after the installation, the numbers of males and females injured were approximately the same.

Type of intersection. Two types of intersections were encountered in the analysis of traffic accident data (**Table 2**). Most intersections (72 percent) at which pedestrian accidents occurred were 4-leg intersections in which two streets crossed each other and continued onward. There were 15 intersections of the other type, called a 3-leg or "T" intersection, in which one of the streets ended at the intersection. Thirty-seven (86 percent) of the forty-three 4-leg intersections were regular, that is, the streets crossed at right angles. The other six were skewed, nonperpendicular crossings. For the 3-leg intersections, 12 were regular and three were skewed.

Twenty-one (49 percent) of the 4-leg intersections had accidents during the period studied, and six (40 percent) of the 3-leg intersections had accidents. The number of accidents at 4-leg intersections decreased by one after the installation of the audible pedestrian signals. However, 3-leg intersections experienced an increase of two accidents. The skewed 3-leg intersections experienced no change in the number of accidents before and after APTS installation.

Further analysis of the available accident data showed that intersection type made little difference in the incidence rate for accidents. Whereas 4-leg intersections comprised 72 percent of the intersections, they experienced 78 percent of the intersections at which there were accidents. Likewise, 3-leg intersections comprised 25 percent of the intersections and experienced 22 percent of the accidents.

Table 2.
Pedestrian accidents by type of intersections

	4-leg intersections			3-leg intersections		
	Total	Regular	Skewed	Total	Regular	Skewed
Number of intersections Total = 60*	43 (72%)	37 (62%)	6 (10%)	15 (25%)	12 (20%)	3 (5%)
Intersections with ped. accidents Total = 27	21 (78%)	19 (70%)	2 (7%)	6 (22%)	4 (15%)	2 (7%)
Intersections without ped. accidents Total = 33	22 (67%)	18 (55%)	4 (12%)	9 (27%)	8 (24%)	1 (3%)
Number of pedestrian accidents Total = 48	40 (83%)	35 (73%)	5 (10%)	8 (17%)	6 (13%)	2 (4%)
Pedestrian accidents before APTS Total = 24	21 (88%)	18 (75%)	3 (13%)	3 (13%)	2 (8%)	1 (5%)
Pedestrian accidents after APTS Total = 24	19 (79%)	17 (71%)	2 (8%)	5 (21%)	4 (17%)	1 (4%)

*Two mid-block pedestrian crossings experienced no accidents.

One important variable—whether any of the injured pedestrians knew of the signal, heard it, and knew what the sound meant—could not be determined from the accident records. Furthermore, the pedestrians involved in these accidents could not be contacted because their names and addresses were confidential. Variables that could be ascertained from the accident data are discussed below.

Vehicle activity. Accidents in which the party at fault was the driver/vehicle were reviewed in terms of when the accidents occurred relative to the installation of the APTS and what was the direction of vehicular travel. Both before and after installation of audible signals, the driver/vehicle was at fault in 60 percent of the accidents, and the vehicle was heading straight ahead through the intersection in half the accidents. There were, however, fewer pedestrian accidents involving cars turning left after the installation of the APTS (25 percent versus 17 percent). This may suggest that fewer pedestrians inappropriately crossed the street during the left turn signal after the installation of audible signals. Two-thirds of the accidents occurred in the first full year following installation of the audible signals, and dropped substantially in the second year.

Pedestrian activity. The accidents in which the pedestrian was at fault were reviewed with respect to the type of intersection and how long the APTS had been installed. Following installation, two-thirds of the accidents occurred in the first full year after the audible signals were installed; the remainder occurred in the year of installation (**Table 3**). In the second year after APTS installation, none of the accidents were caused by the pedestrian. Such a trend may indicate that pedestrians became acquainted with the signal over a period of time before they paid attention to them or understood what the sounds meant.

Prevalence of accidents. The number of accidents per intersection was reviewed. Half (13) of the intersections experienced only one accident, seven intersections had two accidents each, and another seven had three accidents each. The accident records seemed to indicate that fewer pedestrian accidents occurred at the more dangerous intersections (i.e., intersections previously experiencing two or more pedestrian accidents) following APTS installation.

Weather and lighting conditions. The weather and lighting conditions were reviewed for those pedestrian accidents which occurred after the installation of audible signals. Where the driver/vehicle was at fault, the weather was clear in 60 percent of the accidents; and where the pedestrian was at fault, it was clear in 89 percent of the accidents. However, nearly 45 percent of the accidents where the pedestrian was at fault occurred when it was dark.

Table 3.
Party at fault in pedestrian accidents

Time of APTS installation	Party at Fault		
	Vehicle	Pedestrian	Unknown
Before installation of APTS			
2 calendar years before	13	9	2
After installation of APTS			
Calendar year of installation	3	3	—
1st calendar year after	10	6	—
2nd calendar year after	2	0	—
Subtotals for installation	15	9	—

Vehicle traffic volumes. In terms of vehicle traffic volume, two-third of the intersections experiencing pedestrian accidents ranked at the top for traffic volume. These intersections accounted for 83 percent of the accidents before installation of audible signals but only 58 percent after installation. This suggests that audible signals can reduce the number of pedestrians accidents at intersections with high traffic volumes.

REACTIONS TO AUDIBLE PEDESTRIAN TRAFFIC SIGNALS

Installation of audible signals has not been without controversy (4,10,12) with regards to whether they helped persons with vision disabilities to cross streets. Believing that the signals do help, the American Council of the Blind supports their installation. On the other hand, the National Federation of the Blind reports that the signals are greatly disliked by blind persons who do not find them useful (12). One consequence of this disagreement is hesitancy by some public official to continue installing them.

To determine the breadth of the agreement with either points of view, surveys were distributed to: 1) organizations providing services to persons with vision impairments and to older adults; 2) counselors of blind and vision impaired students at educational institutions; and, 3) orientation and mobility instructors who teach blind persons to travel using the long cane technique. The purpose of the surveys was to ascertain how clients and students felt about the usefulness and desirability of the audible signals. Visually impaired persons themselves could not be readily contacted because their names and addresses were confidential.

Survey results from organizations that serve the elderly and persons with visual impairments

Eighty organizations in the United States and one in Canada were surveyed about their knowledge and appraisal of audible traffic signals. Survey responses were received from 37 organizations.

- Thirty-three organizations (89 percent) were familiar with audible signals. Organizations serving blind persons and schools with programs for blind students had higher response rates than organizations serving senior citizens.
- Almost all (94 percent) respondents were aware of audible signals. Four of five reported that their clients had favorable experiences with APTS; the remainder reported that their clients had negative experiences.
- Half the organizations have taken a position on audible signals. Of these, 83 percent support the installation of APTS because the signals were or could be helpful to their clients.
- Almost half the responding organizations have advocated for the installation of audible signals, and 71 percent of these efforts have proven successful.
- Respondents reported that their blind members liked the signals and found them useful at complex intersections (e.g., 5-legged intersections, divided roads with left turn lanes, etc.) and where there is light parallel traffic flow.
- Respondents noted a need for standardization of equipment and for training in its use.

Survey results from educators and school counselors

Because APTS have been installed near educational institutions, all California school districts and institutions of higher education known to have audible signals within their communities were contacted by phone. The instructors and counselors were invited to express their opinions and experiences with the signals and describe student reactions to them. Survey findings from 17 California educational institutions were as follows.

- Blind students were trained to use audible signals at about 12 years of age, and their teachers reported that students found the signals easy to use.
- Counselors at colleges and universities with audible signals nearby reported that their students liked the signals and found them useful. No counselor reported receiving criticism of the signals from any of their blind students.
- Half the respondents volunteered that their sighted students also found the signals useful, as did blind professors at two of the surveyed institutions.

Survey results from instructors of blind person

The entire membership of the California Association of Orientation and Mobility Specialists was surveyed by mail. The instructors reported that they had discussed audible signals at their meetings and that many members have had experience in teaching blind clients how to use them. Of the 45 members surveyed, 27 responded; only one respondent was not personally familiar with audible signals. Major findings from this survey were as follows.

- Most clients found the signals easy to use. Nearly 60 percent of the respondents reported that their clients usually or always found the signals helpful.
- More than half of the clients learned to use the audible signals after only one to two training sessions.
- Clients especially liked knowing when the WALK signal was illuminated, enabling them to feel safer when crossing a street.
- Clients had trouble remembering which sounds designated the north-south and east-west directions of travel.
- Clients had difficulty locating the push button to actuate the pedestrian WALK signal.
- Clients sometimes became disoriented in crossing streets when the audible sound stopped as the WALK light changed to a flashing red DON'T WALK light.
- Some clients had difficulty hearing the signal over the traffic noise, properly aligning themselves at the corner, or became excessively dependent on the audible signal.

CONCLUSIONS AND RECOMMENDATIONS

Audible pedestrian traffic signals have been installed with the intention of helping visually impaired persons cross streets with greater safety and confidence. This article has examined the prevalence of these devices in North America and the impact of the Nagoya/Traconex APTS on accident rates in nine California cities. Reactions and acceptance of APTs were ascertained by surveying knowledgeable professionals and institutions that interact regularly with visually impaired and elderly persons. The survey results and accident data appear to support the following conclusions and recommendations.

- Approximately 300 intersections in the United States and Canada have installed APTS, and many cities are considering installing them in the near future. The Nagoya/Traconex unit is used almost exclusively in the western U.S., while the Sonalert unit by Mallory is more widely used in the eastern U.S.

- Presently, the distribution of APTS is quite uneven, with a vast majority being located in California.
- At intersections where the WALK signal is always operating, traffic engineers have added timers to deactivate the APTS at night and thus avoid disturbing nearby residents.
- Based on the accident data from nine California cities, the audible signals have not yet made a clear impact on pedestrian accident rates at the intersections studied despite generally favorable user responses to such signals.
- For some of the intersections studied, the audible signal appeared to lessen the frequency of inappropriate pedestrian crossings.
- The driver/vehicle was at fault in nearly two-thirds of the accidents so audible signals would not have helped in such cases.
- The pedestrian was at fault in just over one-third of the accidents following APTS installation. These accidents occurred either in the year of APTS installation or during the following year. No pedestrian-caused accidents occurred in the second full year following APTS installation. It appears that APTS may be beneficial to all pedestrians once they become familiar with them. Therefore, public educational programs regarding APTS should accompany their installation.
- Because fewer persons under age 17 were involved in accidents following APTS installation, audible signals may especially help young people to cross streets more safely.
- There is substantial support for audible signals from the surveyed organizations and from professional that provide training services to visually impaired pedestrians.
- Many blind persons, once properly trained, find the audible signals useful in helping them to cross streets safely and with greater confidence. However, a minority of respondents do not want audible signals and believe that they would not be useful to blind pedestrians.
- Programs should be established at senior citizen centers and other similar agencies to instruct older adults, who are developing vision impairments as part of the aging process, in pedestrian safety and how to use audible signals.
- Some visually impaired persons had difficulty finding the traffic light pole and push button that activated the WALK light. Standardizing the location of such poles and their push buttons at corners of intersections would therefore be helpful.

Additional research is needed to address several unanswered questions: What is the impact of the Sonalert and other types of audible signals on pedestrian accident rates? Although the number of pedestrian accidents at the intersections studied did not change, did the installation of APTS reduce the pedestrian accident rate if vehicular and pedestrian traffic volume changes were taken into account? Would expanding the traffic accident rates beyond the 2 years following APTS installation establish a clearer trend as to their impact on pedestrian traffic safety?

ACKNOWLEDGMENTS

This project was supported in part by a grant from the National Institute on Disability and Rehabilitation Research. The suggestions of the reviewers and editor have been appreciated.

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Department of Veterans Affairs

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A Technical Note

Audible pedestrian traffic signals: Part 2. Analysis of sounds emitted

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Abstract—This project evaluated audible pedestrian traffic signals from three perspectives: 1) the patterns of use and the impact of these signals on pedestrian travel; 2) the physical characteristics of the sound emitted by these devices; and, 3) the detection of their emitted sounds in the presence of various traffic noise levels. This paper, the second of three companion articles (2, 3), examines the sounds emitted by the Nagoya/Traconex audible traffic signal, the unit most commonly found in the western United States and almost exclusively in California. The sounds emitted by the north-south and east-west Traconex audible signals were analyzed for their loudness, directionality, frequency spectrum, and temporal characteristics using standard engineering tools including an anechoic chamber, sound level meters, spectrum analyzers, and signal analyzers.

Key words: *audible pedestrian traffic signals, elderly, signaling devices, sonogram, sound dispersion, sound intensity, spectral analysis of sounds, visually impaired.*

INTRODUCTION

In a companion paper (2), the prevalence and impact of audible traffic signals were described. The second phase of this project characterized the sounds emitted by Nagoya/Traconex Audible Pedestrian Traffic Signals (APTS) using standard engineering instruments such as sound meters, frequency analyzers, high fidelity recorder, and an anechoic chamber. As illustrated in **Figure 1**, the Nagoya/Traconex signal is a self-contained, bell-shaped, and weatherproofed unit that attaches via brackets directly to the pedestrian crosswalk signal box. The analysis of sounds from these audible signals consisted of three parts: 1) a spectral analysis of the sounds emitted by the device based on field recordings and anechoic chamber tests; 2) the sound intensity level reaching a pedestrian at an intersection with the Nayoya/Traconex device; and, 3) the directionality of the emitted sound.

The technical analysis focused on the Nagoya/Traconex audible traffic signal for several reasons. First, such units were available to the project for extended testing (3) from Traconex, the manufacturer under license from Nagoya Electric Works of Japan. Second, such units are exclusively used at many intersections in many California cities so that multiple field inspections of installed units and field recordings of their emitted sounds were possible. Third, the analysis of sounds from these devices could be related to their impact (2) on the traffic accident data available from the California Highway Patrol.

METHOD

Recordings of the audible pedestrian traffic signal (APTS) were made at three intersections in San Diego, CA, where audible pedestrian traffic signals had been installed. A high fidelity portable cassette recorder (SONY PMD221) made recordings of the sounds for both directions of travel, north-south and east-west, in December of 1987, during the mid-morning hours of 9:30-11:00 a.m., and during the evening traffic rush hours of 4:00-6:00 p.m. The intersections studied were Mission Boulevard and West Mission

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Figure 1.
The Nagoya Traconex audible pedestrian traffic signal installation for the north-south and east-west directions of travel.

Bay Drive, College Avenue and El Cajon Boulevard, and College Avenue and Montezuma Road. The overall level of traffic noise at these locations was also measured using a hand-held, calibrated sound level meter (Bruel & Kjaer Type 2226).

Sounds from Nagoya/Traconex audible signals

The Nagoya/Traconex audible pedestrian traffic signals emit an electronic "cuckoo" sound for the north-south direction of travel and an electronic "chirp" sound for the east-west direction of travel. The units are designed to emit their respective sounds continuously during the WALK phase of the traffic signal cycle. The APTS emit no sound during the flashing or steady DON'T WALK phases.

The manufacturer (4) lists the following specifications for its north-south "cuckoo" sound: 1) two frequencies are combined to produce the cuckoo sound; 2) the duration of this sound is 0.6 s $\pm$ 20 percent; 3) the frequency base is 1100 Hz $\pm$ 20 percent; and, 4) the frequency deviation is + 120 Hz $\pm$ 20 percent. Specifications for the east-west "chirp" sound are as follows: 1) a continuous frequency variation to produce the chirp sound; 2) the chirp lasting

0.2 s ± 20 percent; 3) a frequency base of 2800 Hz ± 20 percent; and, 4) a frequency deviation of 800 Hz ± 20 percent.

The output for both units is specified to be 90 dB sound pressure level (SPL, re; 20 μ Pa) at 1 m and self-switching between one of two adjustable output levels depending on the ambient noise level. The power requirement of the equipment is 115 AC ± 15 percent, 60 Hz, 3 W.

The weatherproof, bell-shaped units are 16.5 cm (6.5 in) high and 13.3 cm (5.25 in) in diameter and weigh 1.4 kg (3 lbs). They are attached by brackets to the signal box holding the pedestrian traffic signal light (**Figure 1**) and wired in parallel to the WALK or green light indicators. Thus a typical 4-way intersection would have eight pedestrian signal boxes and eight audible signals.

The audible signal equipment used for the laboratory measurements were new Nagoya/Traconex units supplied to the project by the importer, Traconex, Inc. The field measurements and recordings were made on units that were installed on traffic signal poles and that had been in use for about 3 to 4 years.

Field measurements

Physical inspections of the Nagoya/Traconex audible traffic signals installed in various locations throughout San Diego revealed the following:

- The audible traffic signal units were usually mounted on traffic signal poles and attached to their respective walk signals. This location usually meant that the units were 2.5 to 3 m (8 to 10 ft) away from the ears of a typical pedestrian standing on the corner waiting to cross the intersection.
- The north-south signal emitted a *longer* cuckoo sound pattern that repeated a minimum of *four* times during the WALK portion of the light cycle.
- The east-west signal emitted a *shorter* chirp sound pattern that repeated a minimum of *seven* times during the WALK portion of the light cycle.

The field recordings and field sound meter measurements revealed the following:

- The sound intensity directly in front of the device (3 cm away) was 105-110 dBA SPL (A weighted).
- The traffic noise at these intersections varied between 55 dBA and 85 dBA SPL.
- The audible traffic signals momentarily increased their sound outputs by approximately 5 dB (as claimed by the manufacturer) when the traffic noise was extra loud (over 80 dBA SPL), such as with the passing of a diesel truck or bus.

Spectral analysis of field recordings

Using a high fidelity portable cassette recorder (SONY PMD221), the sounds emitted by audible traffic signals were recorded from where a pedestrian would stand while waiting to cross the intersection. Subsequent spectral analysis of the recorded sounds using a Scientific Atlanta SD300 signal analyzer and the Digital Sona-Graph (Model 7800) by Kay Elemetrics Corporation, Pine Brook, NJ, revealed the following:

- The sonograms for both the north-south and east-west signals, recorded from where a pedestrian would stand while waiting for the corresponding traffic light to signal its WALK message, are shown in **Figure 2** and **Figure 3** respectively. The horizontal axis represents time (100 ms per major division) while the vertical axis represents sound level intensity (5 dB per division) or frequency (500 Hz per division). The top trace in the figures shows that the emitted sound was 4 to 5 dB above the background noise level.
- A frequency analysis of the cuckoo sound from the north-south signal (**Figure 2**) shows major frequency peaks at approximately 950, 1950, 2875, 3825, and 4725 Hz. This sound pattern lasts about 400 ms, repeats once every 1.5 s, and has a characteristic sound of an electronic "cuckoo" produced by incrementally changing from one frequency to another (1250 Hz to 950 Hz) with harmonics as high as 6000 Hz.
- A frequency analysis of the chirp sound from the east-west signal (**Figure 3**) shows major frequency peaks at approximately 2100 and 6300 Hz. The east-west chirp signal has a continuous variation in frequency fundamentals from 2600 to 1500 Hz with harmonics up to 7000 Hz. This sound pattern lasts about 140 ms and repeats at 1 s intervals.
- Spectral analysis of the traffic noise revealed wide-band noise with frequency components from 6 Hz to 7000 Hz with most of the acoustic energy below 1000 Hz.

Laboratory measurements of the sound intensity patterns

Careful and multiple laboratory measurements of the sound intensity and sound directionality of the new Nagoya/Traconex units were made inside an anechoic chamber located on the San Diego State University campus. A calibrated cardioid microphone was placed 1 m away from the signal and oriented from +90 to −90 degrees in 30-degree increments about the main axis (0 degrees) of the sound source. The sound intensities (in dB SPL, re: 20 μ Pa) of the north-south and east-west audible sig-

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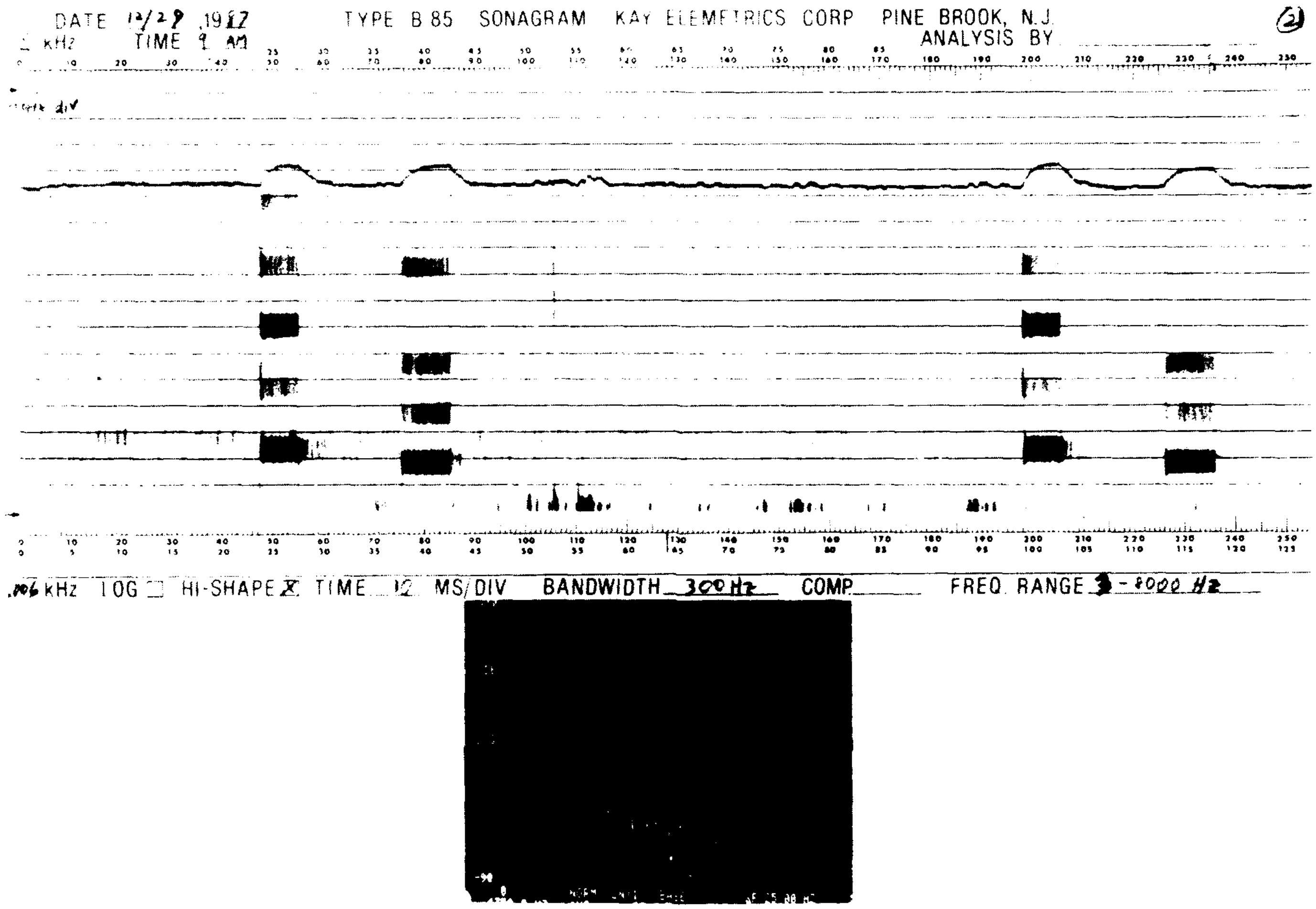

Figure 2.
Sonogram (above) and frequency spectrum (below) of the north-south "cuckoo" sound. The horizontal scale is 10 ms per division, and the vertical scale is 500 Hz (or 5 dB) per division.

nals were measured with their volume controls at the center (or medium) and maximum settings.

With the volume control of the north-south cuckoo signal set at the *center* position of its adjustments range, the output sound intensity varied from 76 dB at 0 degrees (straight ahead) to 70 dB at ± 90 degrees (**Figure 4**). When the volume control of the north-south cuckoo signal was set for *maximum* output, the signal's sound intensity at 0 degrees surprisingly stayed the same (76 dB). However, the sound intensity exhibited slightly less directionality, showing a drop of just 2 dB at the ± 90 degree extremes.

With the volume control of the east-west chirp signal set at the *center* position of its adjustment range, the output sound intensity varied from 71 dB to 0 degrees to 59 dB at ± 90 degrees (**Figure 5**). With the volume control of the east-west chirp signal set for *maximum* output, the signal's sounds intensity increased noticeably (about 6 dB). The chirp's intensity was 77 dB at 0 degrees and 65 dB at ± 90 degrees.

SUMMARY

The field recordings and anechoic chamber tests of the Nagoya/Traconex audible pedestrian traffic signals revealed the following:

- Sound chamber evaluations of the Nagoya/Traconex audible pedestrian traffic signal indicate that the sound emitted by the north-south unit was less directional than the sound emitted by the east-west unit. However, their sound patterns (**Figure 4** and **Figure 5**) were not so highly directional that precise aiming of the units during field installations would be necessary.

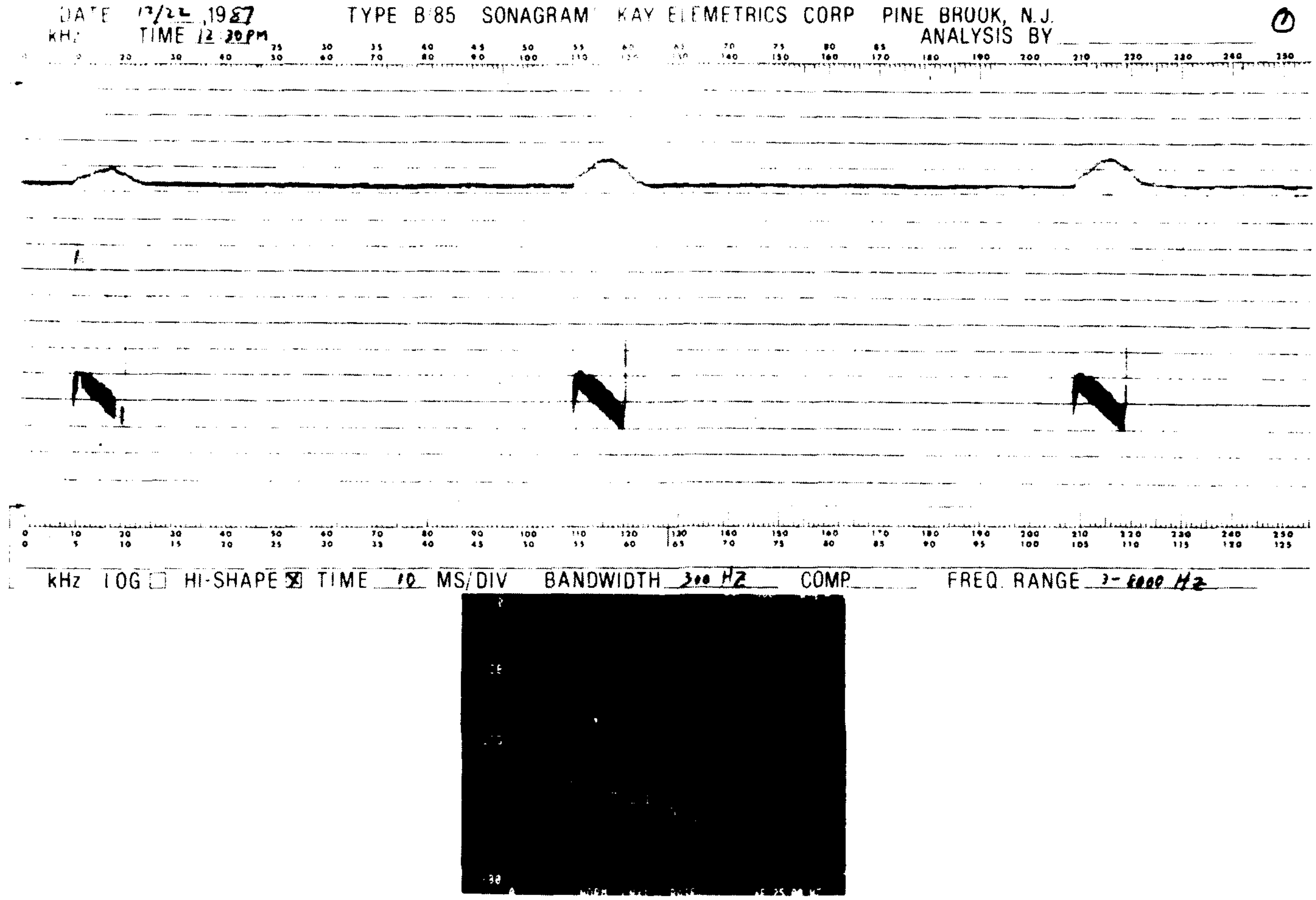

Figure 3.
Sonogram (above) and frequency spectrum (below) of the east-west "chirp" sound. The horizontal scale is 10 ms per division, and the vertical scale is 500 Hz (or 5 dB) per division.

- Similar volume control settings on these units may not yield similar outputs. Therefore, accurate on-site settings of the audible signal's output must be made with the aid of a calibrated sound level meter.
- The east-west chirp sound has much higher frequency components than the north-south cuckoo sound. Thus the east-west chirp probably would be more easily detected than the north-south cuckoo in the presence of the broadband, low frequency traffic noise (3).
- Assuming that the Traconex APTS's maximum sound pressure output would increase about 5-8 dB in the presence of loud environmental noise, the test unit met the factory listed specifications within reasonable manufacturing tolerances.

ACKNOWLEDGMENT

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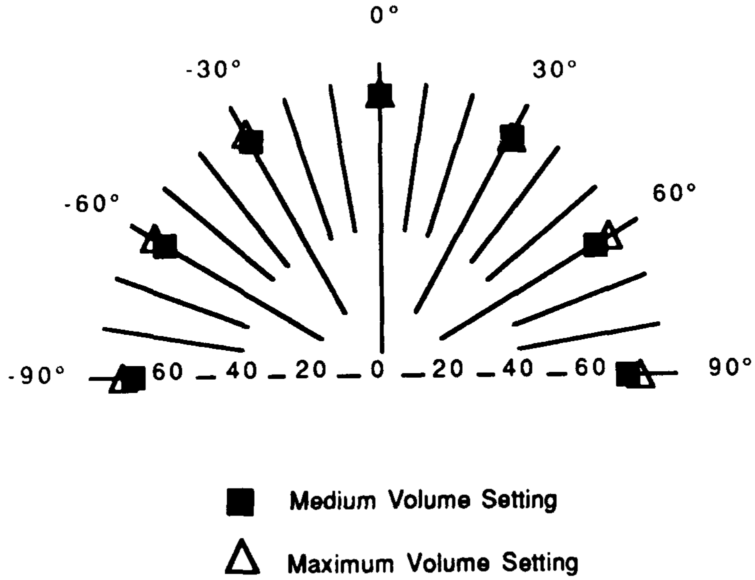
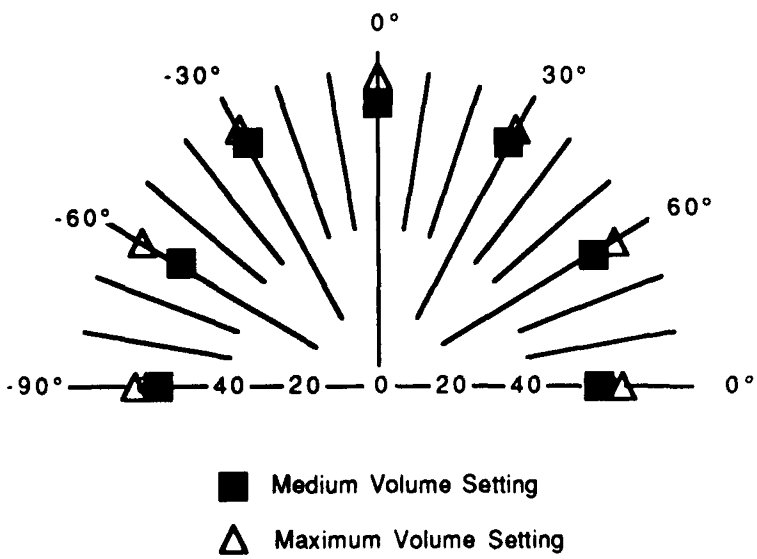
NORTH-SOUTH "CUCKOO" SOUND PATTERNS



Orientation	dB SPL (Med.)	dB SPL (Max.)
-90°	71	74
-60°	72	75
-30°	74	76
0°	76	76
30°	75	75
60°	71	75
90°	70	74

Figure 4.
North-south "cuckoo" sound dispersion patterns at medium and maximum volume settings.

EAST-WEST "CHIRP" SOUND PATTERNS



Orientation	dB SPL (Med.)	dB SPL (Max.)
-90°	59	65
-60°	61	73
-30°	70	74
0°	71	77
30°	70	74
60°	66	73
90°	59	65

Figure 5.
East-west "chirp" sound dispersion patterns at medium and maximum volume settings.

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A Technical Note

Audible pedestrian traffic signals: Part 3. Detectability

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Abstract—This project (10) evaluated audible pedestrian traffic signals (APTS) from three perspectives: 1) the patterns of use and the impact of these signals on pedestrian travel; 2) the physical characteristics of the sound emitted by the Nagoya/Traconex APTS; and, 3) the detectability of the sounds emitted by this brand of APTS. This paper, the last of three companion articles (13,14), describes the detectability of the sounds emitted by the Nagoya/Traconex audible traffic signal, the unit most commonly found in the western United States and almost exclusively in California. To determine detectability, three groups of subjects with normal hearing—young sighted adults (controls), elderly sighted adults, and elderly blind adults—participated in an audiological study. Auditory stimuli, which consisted of APTS sounds embedded in various levels of interfering traffic noise, were presented to subjects seated inside a double-walled sound-treated chamber. The subjects were instructed to press down on a response button as soon as they heard the audible pedestrian traffic signal. The percentage of correct detections determined the absolute detectability of APTS under various S/N ratios. The subjects' speed of response indicated how quickly a pedestrian might begin to cross the intersection upon hearing the APTS.

Key words: *audible pedestrian traffic signals, audiological tests, detectability, elderly, signal to noise ratio, speed of response, traffic noise, visually impaired.*

BACKGROUND

In order for audible pedestrian traffic signals to be effective, the sounds they produce must be easily heard so that an appropriate response can be elicited. A review of the literature examined several issues that bear on this stimulus-response behavior. These issues include auditory sensitivity, effects of masking due to traffic noise, signal type and duration, effects of age, and the implications of signal detection theory with respect to the desired response.

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Auditory sensitivity and traffic noise interference

Auditory threshold for the range of frequencies heard by the human ear (20 to 20,000 Hz) is best between 2000 to 5000 Hz. This means that audible traffic signals will be the most readily heard if their sounds contain frequency components within this range, even if there is low frequency interference such as that from traffic noise. In their spectral analysis of traffic noise, Welsh and Blasch (16) found 78 dB of sound pressure level (SPL) at 125 Hz and only 45 dB SPL at 12,500 Hz when the overall traffic level was 84 dB SPL.

Although most of the sound energy in traffic noise occurs in frequencies below 2000 Hz, low frequency sounds can nevertheless effectively mask signals in the 2000 to 5000 Hz region if the intensity of the masker is sufficiently strong (80 to 100 dB SPL) (6). Recorded traffic noise samples at intersections studied in this project ranged from 55 dB to 85 dB SPL. Therefore, low frequency masking of the audible pedestrian signal is possible.

Signal type

Besides the level of background noise, spectral complexity of the signal also affects its detectability in noise. Signals that are complex (i.e., ones containing a variety of frequencies at different intensities) are easier to detect

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than a pure tone (3,8). This finding implies that maximally detectable audible traffic signals should emit multiple frequencies.

Signal frequency and duration

Both the frequency and duration of the signal also affect its detectability. Detectability of signals in noise steadily worsens with signal durations shorter than 500 ms (7). Furthermore, low frequencies (< 1000 Hz) require more intensity or longer durations than high frequencies (> 3000 Hz) to be equivalently detectable (9). Therefore, audible traffic signals should be at least 500 ms in duration and contain frequency components above 3000 Hz.

Effects of age

Audible traffic signals are intended to help blind and sight impaired pedestrians, many of whom are elderly. Smith and Sethi (11) found a slowing of brain wave activity in their healthy elderly subjects and a delay in central reaction times based on their longer time for recognition and slower response speeds. The Committee on Hearing Bioacoustics and Biomechanics reported that a slowdown of both sensory motor and mental processes occurred with aging even in normal hearing older subjects (4). Blackington (1) found that elderly subjects with normal hearing were also more susceptible than their young normal hearing counterparts to the effects of masking noise when that noise precedes (forward masking) the target signal. The groups, however, performed similarly when the signal and the masker were presented simultaneously. Both simultaneous and forward masking conditions due to traffic noise exist for the older pedestrian trying to detect the audible traffic signal at a typical intersection.

Theory of signal detection

Detectability of signals depends not only on the relative intensities of the traffic noise and signal, but also on how different the signal is from the noise and on the costs associated with making an *incorrect* decision. The theory of signal detection (TSD) directly addresses this issue. TSD makes the distinction between what the subject actually hears (auditory sensitivity) and the manner in which he/she responds. The response reflects not only a subject's auditory sensitivity, but also the bias and response criteria assumed by the subject (12).

TSD states that a subject's criteria for making a response to a signal imbedded in noise are based on three factors: 1) the probability of there being a signal plus noise versus noise alone; 2) the amount of overlap between conditions of signal plus noise and noise alone; and, 3) the values and costs associated with the outcome of either a walk or don't walk decision. In short, a subject's response to an audible traffic signal is strongly dependent on auditory sensitivity, the separation between the noise alone and signal plus noise probability distributions, the ability of the auditory system to make use of this separation, and the subject's relative value criteria for correct and incorrect responses.

Response time has also been examined in the context of TSD (7). It was observed that as signal to noise (S/N) ratios became more negative (difficult) and fell below the decision criterion, "no" decisions (i.e., signal was not heard) would become more rapid, while less negative (easier) S/N ratios above the decision criterion resulted in more rapid "yes" decisions (i.e., signal was heard). Emmerich, Gray, Watson, and Tanis (5) found that a listener's response latency shortened as he/she became more confident that a signal in noise was heard. Similarly, as the listener became more confident that there was not a signal (noise only), his/her response speed for a "no" response increased. These findings are relevant to the blind pedestrian who needs to be highly confident that the audible traffic signal was present before he/she would begin crossing the street. Thus, decision time or response latency would be expected to lengthen under conditions of poorer S/N ratios.

With the above as background, an audiological study (2) was conducted to: 1) determine the relative detectability of the north-south (cuckoo) and east-west (chirp) APTS in the presence of various levels of background traffic noise (S/N ratios of −5 to −30 dB); and, 2) compare the response times to these signals from three subject groups: young normal-sighted normal-hearing, elderly normal-sighted normal-hearing, and elderly normal-hearing blind subjects. Audible pedestrian traffic signals made by Nagoya/Traconex (15) were chosen for this audiological study because they are almost exclusively used in California and because the manufacturer made them available for extended use by the research team.

METHODOLOGY

Seven young normal-sighted adults ages 22 to 35 years (mean of 29.1 years), seven elderly normal-sighted adults ages 61-78 years (mean of 67.6 years), and five legally blind adults ages 62 to 84 years (mean of 73 years) participated in this study. "Normal sighted" meant the ability to read-

ily see the WALK/DON'T WALK traffic signal from across the street. "Normal hearing" meant pure tone air conduction thresholds of 25 dB HL or better for frequencies between 500 and 4000 Hz (ANS, 1969). All subjects had unremarkable medical histories.

Stimulus parameters

Field recordings of the sound pressure levels of the audible traffic signals and traffic noise levels were made at three busy intersections that had audible pedestrian traffic signals in place (14). The decibel (dB) sound pressure levels (SPLs) for the signals were 105 to 110 dB SPL (A weighted) measured 3 cm in front of the device. The traffic noise ranged from 55 dBA to 85 dBA peak impulse, measured at 9:30 to 11:00 a.m. and during the evening rush hours of 4:00 to 6:00 p.m.

The north-south cuckoo signal contained major frequency peaks at approximately 950, 1950, 2875, 3825, and 4725 Hz. The sound pattern lasted about 400 ms, repeated once every 1.5 s, and had a characteristic sound of an electronic "cuckoo" produced by incrementally changing from one primary frequency to another (1250 Hz to 950 Hz). In contract, the east-west chirp signal contained major frequency peaks at approximately 2100 and 6300 Hz. The electronic chirp was produced by a continuous variation in frequency fundamentals from 2600 to 1500 Hz with harmonics up to 7000 Hz. This sound pattern lasted about 140 ms and repeated at 1-second intervals. Spectral analysis of the traffic noise revealed wide-band noise with frequency components from 6 Hz to 7000 Hz. Most of the acoustic energy, however, was below 1000 Hz (14).

Stimulus generation

Special tape recordings of the traffic signals and traffic noise were made for the audiological study. To obtain a clear recording of the signals without background noise, the sounds emitted by north-south and east-west Nagoya/Traconex audible pedestrian traffic signals were tape-recorded in a double-walled sound treated test chamber using a Sharp tape recorder. To obtain a somewhat constant source of traffic noise, a continuous loop of tape (10 s) was made by retaping on automatic gain the earlier field recordings of the wide-band traffic noise.

The continuous loop traffic noise tape, played back on a Sharp tape recorder, was channeled into a Lafayette Instrument module. The tape recorder's output was fed directly into the shaped rise/fall audio switch (ANL-913), and then onto the 600 ohm attenuator (ANL-917). The traffic noise was shaped with a rise/fall time of 10 ms, attenuated, and fed into the audio amplifier/mixer (ANL-914). The traffic signal tapes were played back on a high fidelity Marantz tape recorder (Model PMD 221), channeled into a second attenuator (ANL-917), and then fed to the audio amplifier/mixer (ANL-914) where the traffic noise and signals were mixed (**Figure 1**).

After mixing, the signal and noise stimuli were fed into a Madsen clinical audiometer (OB822). The audiometer was set at a fixed output intensity level of 65 dB HL, and the sounds were presented binaurally via standard TDH-39 earphones. The traffic signal attenuator (ANL-917) was adjusted to achieve presentations at the following S/N ratios: −5, −10, −15, −20, −25, −30, −35 dB.

Custom designed software for an Apple II Plus computer started each presentation by turning on the traffic noise; controlled the length of the traffic noise presentation (10 s); generated a random time element before starting the APTS signal during the traffic noise presentation; and recorded the subjects' response times to the traffic signals.

Calibration of the Bruel and Kjaer sound level meter (Type 2226) was verified using a Quest CA22 sound calibrator. The level of the traffic noise and signals were calibrated into dB SPL values through a Madsen clinical audiometer (OB822) at the earphone using a Bruel and Kjaer sound level meter (Type 2203). At the onset of testing, intensity levels for both the signals and noise were adjusted to peak at zero dB on the V.U. meter of the audiometer.

Test procedures

All testing was performed with the subjects comfortably seated inside a double-walled sound treated test chamber. The auditory stimulus was presented binaurally via TDH-39 earphones housed in MX/AR cushions. The subjects were instructed to press down on a hand-held response button as soon as they heard the audible pedestrian traffic signal. For each trial, a "get ready" warning light illuminated, and then 10 s of traffic noise began. Following a random delay, the cuckoo or chirp signal began to sound. The traffic noise sound level was set at 65 dB HL while the traffic signal presentation level varied in intensity to achieve the desired S/N ratios. For each stimulus trial, a series of 3 cuckoos or 4 chirps were present during the 10 s of traffic noise.

All seven signal to noise ratios were presented to each subject during every 2-hour test session. The entire procedure consisted of two trials of five presentations each, heard at each of the seven S/N ratios. To familiarize

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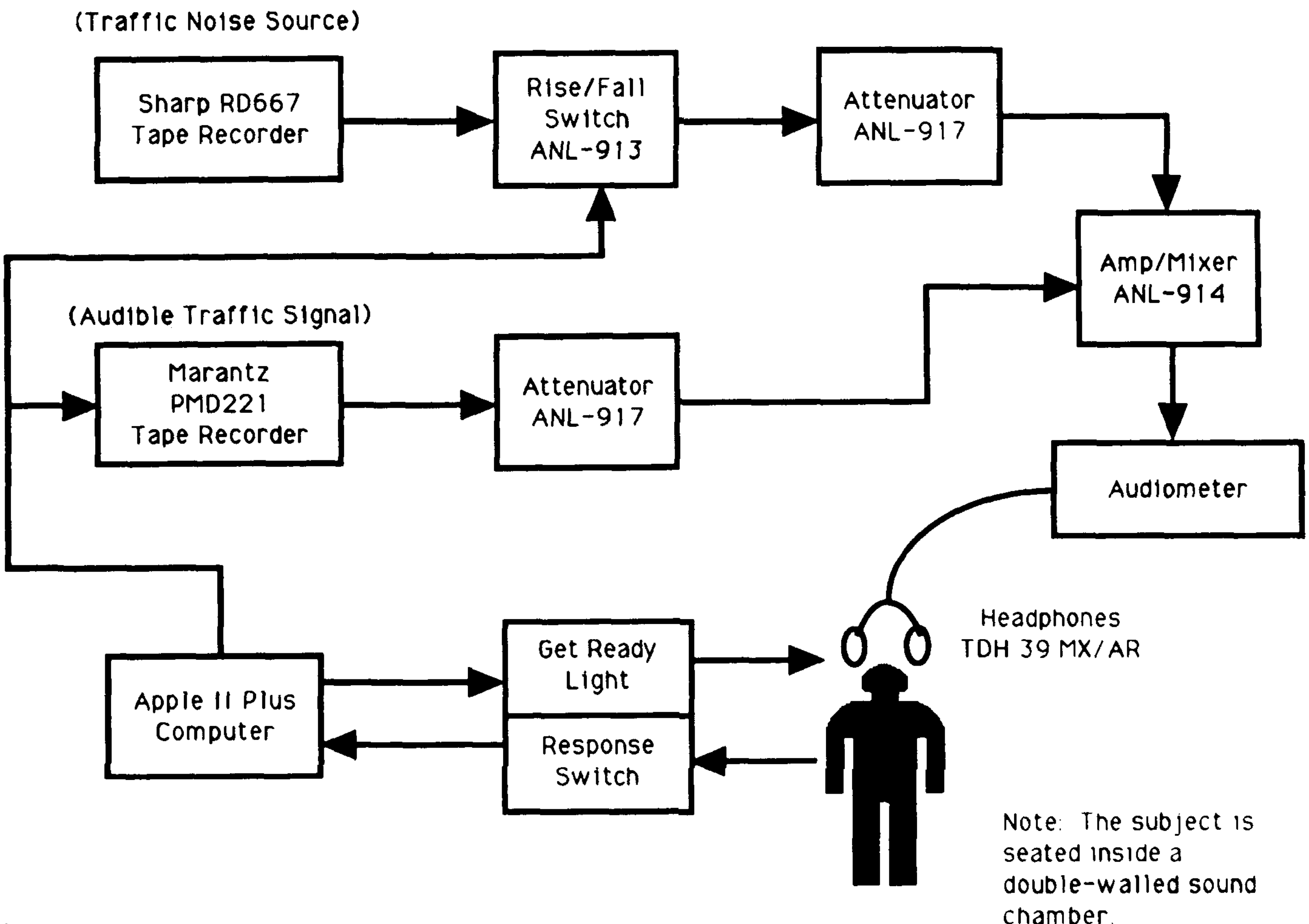

Figure 1.
Block diagram of the audiological test set-up.

the subjects with the task, testing began with a practice trial (five presentations of the auditory stimulus) at zero dB S/N ratio. The two APTS signals (cuckoo and chirp) were tested separately with the presentation order of the S/N ratios randomized.

RESULTS

For each stimulus presentation, a response time was recorded in milliseconds. If the subject did not hear the signal and therefore did not respond, a "no response" score was recorded by the computer. The five presentations in each trial and for each stimulus condition (e.g., −5 dB S/N ratio, trial 1, north-south signal) were averaged together to obtain the response time for that test condition. Detectability of the two sounds under various S/N ratios was calculated by converting the raw data into percentage correct response scores. The response data from all the subjects and all test conditions were then prepared for analysis by the Statistical Package for the Social Sciences (SPSS/PC+) computer program (2). The criterion for statistical significance was set at $p \leq 0.05$.

Percent correct detection

A multivariate analysis of variance (MANOVA) with repeated measures determined if there were statistically significant differences in percentage of correct detection as a function of group membership, signal type, S/N ratio, and trial number. One-way analyses of variance were applied to the significant components obtained from the MANOVA. The Scheffé post hoc procedure was applied to all significant ($p \leq 0.05$) results. The bar graphs in **Figure 2** and **Figure 3** depict the average percent correct detection of both signals under the tested S/N ratios.

The multivariate analysis of variance (summarized in **Table 1**) revealed that:

- For both the cuckoo and chirp signals, the elderly sighted group had noticeably poorer signal detection at the more difficult S/N ratios (−20 through −30 dB). At the −35 dB S/N ratio, all groups achieved 0 percent detection.
- For the north-south cuckoo signal, the elderly sighted group had a significantly poorer ($p \leq 0.05$) detection rate at the −25 dB S/N ratio than either the young sighted or elderly blind groups.
- For the east-west chirp signal, all three test groups were similar (i.e., not significantly different) in their ability to detect this signal.
- One hundred percent correct detection was achieved by all subject groups for S/N ratios of −15 dB or better.

Response time

A second multivariate analysis of variance (MANOVA) with repeated measures determined if there were statistically significant differences in response time as a function of group membership, signal types, S/N ratio, and trials. Due to the lack of responses at −30 dB and −35 dB S/N ratio, they were excluded from the statistical analysis. The Scheffé post hoc procedure was performed on all significant ($p \leq 0.05$) results. **Table 2** lists the average response times of the three groups, with means and standard deviations for each stimulus condition.

The multivariate analysis of subjects' response times (summarized in **Table 3**) revealed that the elderly sighted group had significantly longer response times than the young sighted group and elderly blind group at −20 and −25 dB S/N ratios for the north-south signal. The performance of the elderly sighted group was significantly poorer on their first listening trial. For the east-west signal, there were no significant differences between the groups under any of the S/N ratios studied.

Table 1.
Summary of multivariate analysis of variance (MANOVA) with repeated measures for percent correct detection as a function of group, signal type, S/N ratio, and trials. (Abstracted from Brand [2])

Source of variance	SS	df	MS	F	F prob.
Between Subjects					
Group	1.22	2	0.61	5.15	0.019*
Error	1.90	16	0.12		
Within Subjects					
Signal type	0.13	1	0.13	4.38	0.053
Signal × group	0.22	2	0.11	3.56	0.052
Error	0.48	16	0.03		
Trial	0.05	1	0.05	3.61	0.075
Trial × group	0.00	2	0.00	0.02	0.982
Error	0.21	16	0.01		
S/N ratio	86.83	6	14.47	208.73	0.000**
S/N × group	2.83	12	0.24	3.41	0.000**
Error	6.66	96	0.07		
Signal × trial	0.00	1	0.00	0.01	0.933
Signal × trial × group	0.01	2	0.01	0.67	0.525
Error	0.13	16	0.01		
Signal × S/N ratio	0.27	6	0.05	1.88	0.092
Signal × S/N × group	0.84	12	0.07	2.93	0.002**
Error	2.30	96	0.02		
Trial × S/N ratio	0.10	6	0.02	2.19	0.051
Trial × S/N × group	0.12	12	0.01	1.32	0.221
Error	0.71	96	0.01		
Signal × trial × S/N	0.01	6	0.00	0.36	0.933
Signal × trial × group	0.05	12	0.00	0.82	0.633
Error	0.50	96	0.01		

*p < 0.05
**p < 0.01

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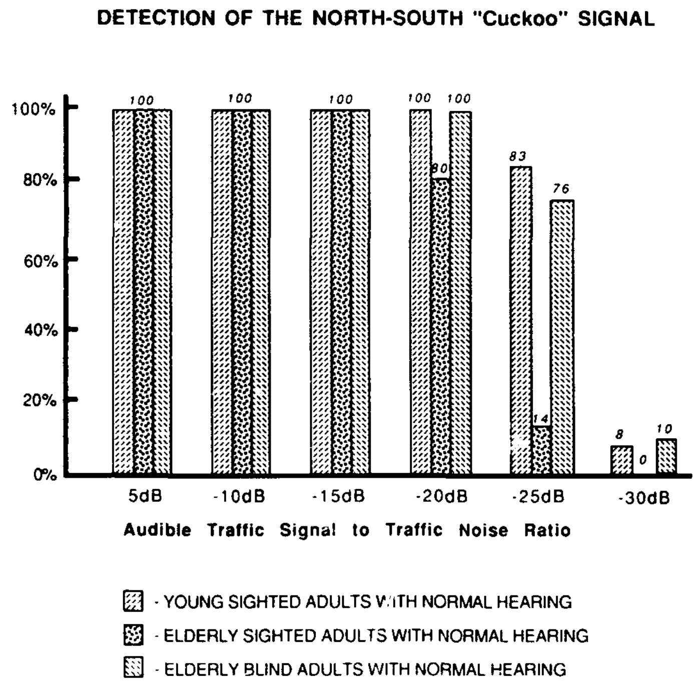

Figure 2.
Percent correct detection of the north-south "cuckoo" signal under various S/N ratios for all three test groups.

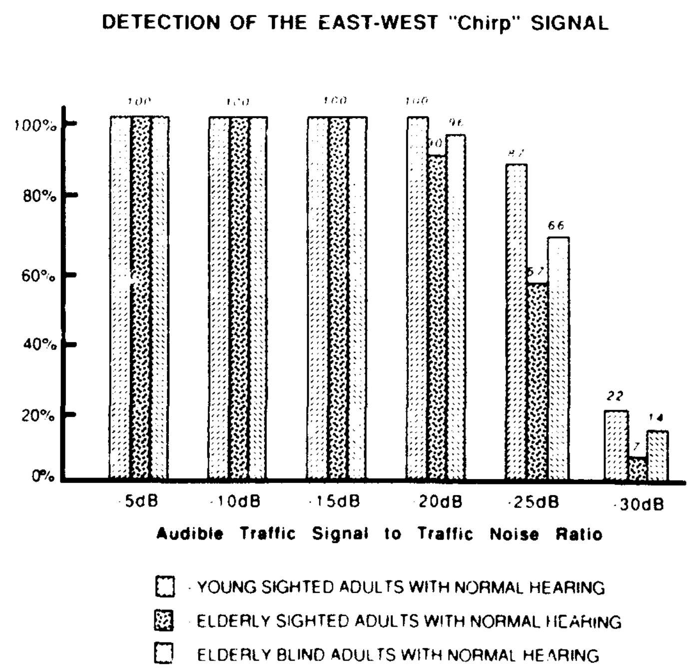

Figure 3.
Percent correct detection of the east-west "chirp" signal under various S/N ratios for all three test groups.

DISCUSSION

The blind-elderly group seemed more resistant to the effects of traffic noise on their detection of and reaction time to both the north-south and east-west audible traffic signals. Their performance was similar to that of the young sighted controls for all stimulus-response conditions. Although all subject groups exhibited longer reaction time as the S/N ratio worsened, only the elderly sighted subjects were significantly poorer than the other two groups at detecting the north-south signal at the more difficult S/N ratios (−20 and −25 dB). The elderly sighted group also required a significantly longer time to react to the appearance of the north-south signal at these S/N ratios in comparison to the other two groups. Because blind persons depend so much on their auditory sense, the elderly blind subject group performed better than their sighted counterparts, particularly when both groups were asked to detect the north-south cuckoo in heavy traffic noise (−20 to −30 dB S/N ratios).

For all three test groups, the electronic chirp was more easily detected than the cuckoo. The reason for this outcome is because the chirp was very spectrally different from the low frequency traffic noise. The chirp had frequencies in the 2000 to 7000 Hz range whereas most of the traffic noise was below 1000 Hz. The chirp was also temporally continuous for about 140 ms whereas the longer cuckoo consists of two quick bursts of low frequency (950 to 1250 Hz) sounds, each of which lasts less than 90 ms (14).

This controlled study occurred in a safe laboratory environment without the danger associated with crossing a street. Based on the theory of signal detection, these same test subjects would reset their decision criteria so that their response times to the "chirp" or "cuckoo" would be longer under the same S/N ratios. Their response times for a "yes" decision would be especially longer as detection becomes more difficult (5,7,12).

Many older pedestrians have bilateral sensory-neural hearing loss above 2000 to 3000 Hz. For them, the detection of the chirp signal will be more difficult due to its higher frequencies and short duration. Thus, even poorer performance under most S/N ratios in terms of response latency and accurate detection of the audible signal can be expected for people with high frequency hearing loss.

Table 2.

Average response times (in seconds) to the north-south and east-west signals with means and standard deviations for the young sighted, elderly sighted, and elderly blind groups, for each of the S/N ratios. (Abstracted from Brand [2])

S/N ratio	Young sighted mean	Young sighted SD	Elderly sighted mean	Elderly sighted SD	Elderly blind mean	Elderly blind SD
North-south						
−5 dB	0.44	0.16	0.45	0.18	0.50	0.17
−10 dB	0.44	0.11	0.46	0.15	0.63	0.25
−15 dB	0.44	0.12	0.56	0.18	0.60	0.28
−20 dB	0.47	0.12	1.11	0.62	0.71	0.30
−25 dB	0.96	0.49	3.11	0.98	1.84	1.14
East-west						
−5 dB	0.27	0.07	0.31	0.06	0.38	0.17
−10 dB	0.32	0.07	0.32	0.10	0.47	0.22
−15 dB	0.34	0.13	0.39	0.07	0.38	0.13
−20 dB	0.48	0.12	0.69	0.60	0.60	0.33
−25 dB	1.34	0.48	1.50	0.49	1.61	0.64

Table 3.

Summary of multivariate analysis of variance (MANOVA) with repeated measures for response time as a function of group, signal type, S/N ratio, and trials. (Abstracted from Brand [2])

Source of variance	SS	df	MS	F	F prob.
Between subjects					
Group	5.90	2	2.95	3.32	0.062
Error	14.20	16	0.89		
Within subjects					
Signal type	1.48	1	1.48	8.44	0.010*
Signal × group	3.58	2	1.79	10.23	0.001**
Error	2.80	16	0.18		
Trial	0.09	1	0.09	1.61	0.223
Trial × group	0.90	2	0.45	7.64	0.005**
Error	0.94	16	0.06		
S/N ratio	66.88	4	16.72	88.50	0.000**
S/N × group	5.45	8	0.68	3.60	0.002**
Error	12.11	64	0.19		
Signal × trial	0.85	1	0.85	7.42	0.015*
Signal × trial × group	0.97	2	0.48	4.21	0.034*
Error	1.84	16	0.11		
Signal × S/N ratio	0.05	4	0.01	0.07	0.990
Signal × S/N × group	6.58	8	0.82	5.34	0.000**
Error	9.86	64	0.15		
Trial × S/N ratio	2.96	4	0.74	11.94	0.000**
Trial × S/N × group	2.94	8	0.37	5.94	0.000**
Error	3.96	64	0.06		
Signal × trial × S/N	1.22	4	0.30	3.50	0.012*
Signal × trial × group	1.01	8	0.13	1.45	0.194
Error	5.55	64	0.09		

*p < 0.05
** < 0.01

CONCLUSIONS AND RECOMMENDATIONS

The acoustic characteristics of an optimal audible pedestrian traffic signal must balance the low frequency (below 1000 Hz) masking potential of traffic noise and the high frequency (above 2500 Hz) sensorineural hearing loss common in older adults. Regardless of the acoustic signal, how well one hears will ultimately determine the effectiveness of these signaling devices. Based on analysis of the collected data, the following conclusions and/or recommendations seem to be indicated:

- Both signals should always yield at least a −15 dB S/N ratio in relation to the existing ambient traffic noise level.
- The audiological study results and foregoing literature review suggest that the north-south cuckoo signal be changed to raise the frequency components to include primary frequencies within the 2000 to 3000 Hz range with harmonics as high as 7000 Hz. The current incremental change of the cuckoo from 1250 to 950 Hz is too easily masked by the louder low frequencies contained in the traffic noise and is too temporally similar to incremental patterns of traffic noise (e.g., automobile horns, passing motorcycles, diesel trucks and buses, and engine acceleration/deceleration).
- An audible pedestrian traffic signal should emit a complex array of frequencies temporally continuous and have a duration of 500 ms. A warble tonal complex with fundamental center frequency of 2500 Hz varying between 2000 and 3000 Hz is recommended. The duration of the east-west chirp signal should also be lengthened to 500 ms per presentation.
- For more effective use of APTS, the WALK phase of the traffic signal should be lengthened 1.5 to 2.0 s when the APTS is in use and when there is loud traffic noise. This amount of additional time for the WALK phase would compensate for the greater response latencies caused by poorer S/N ratios and the element of danger associated with crossing a major roadway.

ACKNOWLEDGMENTS

The contributions of Ms. Katrina Brand in collecting and analyzing the audiological data are gratefully acknowledged. This project was supported in part by a grant from the National Institute on Disability and Rehabilitation Research.

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CLINICAL REPORT

Report on the Evaluation of the VA/SEATTLE Ankle

Prepared by the Rehabilitation R&D Evaluation Unit, Rehabilitation Research and Development Service, Department of Veterans Affairs, Baltimore, MD 21202

Abstract—The VA/SEATTLE Ankle was researched and developed by the Prosthetics Research Study (PRS), Seattle, WA under the sponsorship of the Rehabilitation Research and Development Service, Department of Veterans Affairs. This report presents an evaluation of the project conducted by the Rehabilitation R&D Evaluation Unit (REU), the VA Prosthetic and Sensory Aids Service (PSAS), and the PSAS at 13 VA Medical Centers. The ankle is a lightweight lower limb prosthesis of monolithic design weighting 300 g (10.5 oz.) with a maximum pylon length of 15 1/2 in. Energy storage and release and three-axis motion are all accomplished via a specifically designed convolution in the distal region (ankle) of the pylon. This convolution provides rotation (± 5 degrees), inversion/eversion (± 3 degrees) and dorsi/plantar flexion (± 5 degrees). The ankle showed no signs of fracture or wear through laboratory test procedures of 500,000 cycles of the ankle attached to a conventional type SACH foot under a dynamic load. Performance of the VA/SEATTLE Ankle was found to be acceptable and reliable with a wide variety of users and its commercial production was recommended.

Key words: *amputee gait analysis, lightweight limb prostheses, pylon, SACH Foot, VA/SEATTLE Foot.*

INTRODUCTION

Attempts to improve amputee gait to approach normal physiological walking motion of the torso and lower limbs has been a challenge confronted by physicians, prosthetists, engineers, and therapists. The SEATTLE Ankle is one component in the development of a complete functional, lightweight lower limb prosthesis. It replaces the pylon used in conventional endoskeletal limbs and incorporates limited degrees of plantar/dorsiflexion, inversion/eversion and axial rotation.

Subjects included in the evaluation represented a broad range in age, weight, and activity level. The subjects were asked to rate their performance of certain activities while wearing the SEATTLE Ankle as compared to the prosthesis they previously used. Subjects noted improved performance in the ambulatory functions of walking either slow or fast, running, going up or down stairs, going up or down hills, and going over uneven terrain. The group of subjects with the highest activity level showed the greatest improvement rating. Ninety-eight percent of the subjects said that the ankle movement was smooth during all phases of gait. The ankle functioning was essentially noiseless and only three subjects reported functional failures during the evaluation period.

REVIEW OF DEVELOPMENT

The SEATTLE Rotator Shank evolved into a lightweight multi-axis ankle of monolithic construction and is now known as the SEATTLE Ankle. First generation prototypes were built of graphite fabric layed up wet in a semi-rigid polyurethane matrix, with the bias at 45 degrees in flexing regions and omni-directional in stiff regions. Squeeze molds made of epoxy from hard mandrels compressed the layup to a 70 percent fiber/30 percent resin ration by weight. This material has great potential in other

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FIGURE 1

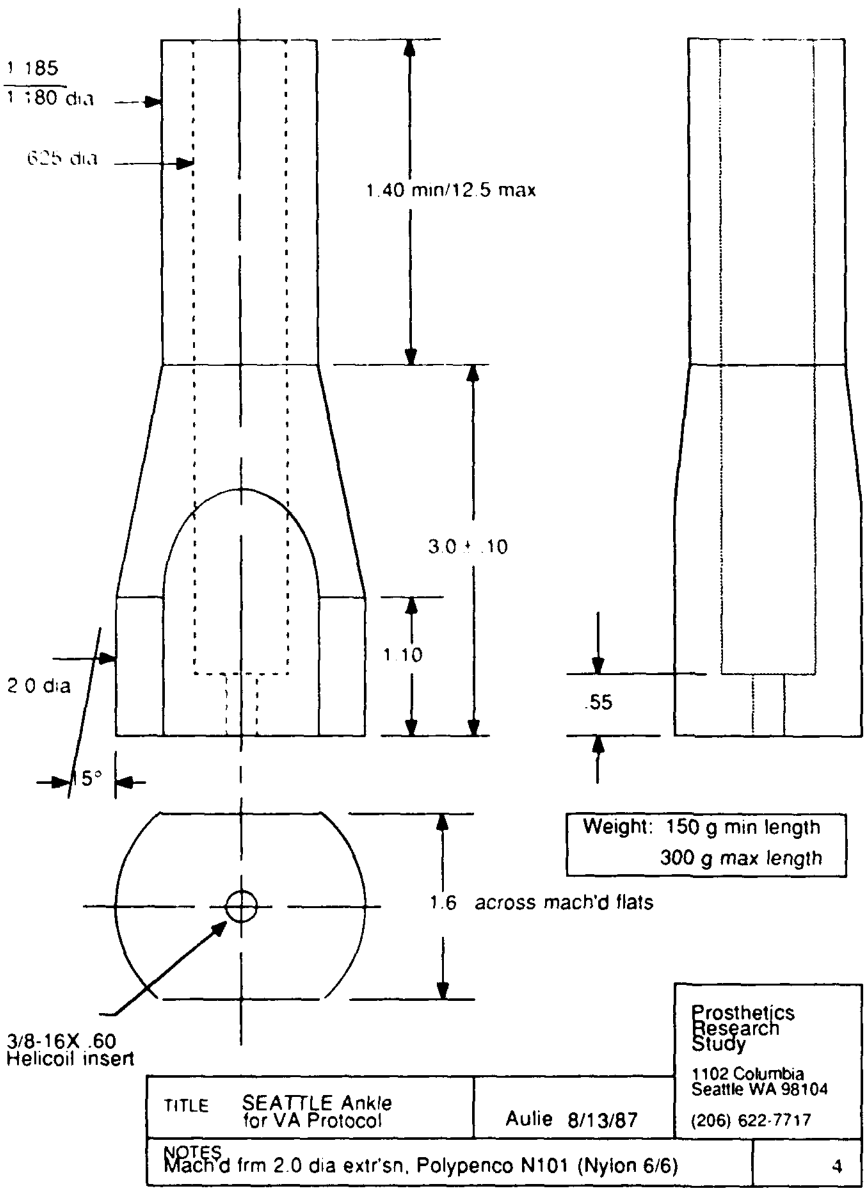

Figure 1.
SEATTLE Ankle for VA protocol.

prosthetic-orthotic applications.

A series of design iterations driven by clinical requirements resulted in a monolithic design with stress levels low enough for application of engineering thermoplastics. The current units are turned by conventional machining methods from 2 in. diameter extruded bars of nylon 6/6. Total manufacturing cost has been reduced drastically and part count reduced to two: the ankle body and a threaded insert for a standard 3/8-16 foot bolt.

Good results of PRS in-house clinical testing were the prime mover in the development of this device, as was the potential for low cost to the end user. Other design constraints included: weight, packaging (e.g., size constraints), corrosion resistance, and loading to 2.5 × a 200 lb. body weight.

At PRS, heavy use testing by eight below-knee and two above-knee amputee subjects resulted in no failures over time periods of up to 16 weeks. Several subjects requested to be returned to the ankle following the test protocol and were fitted for long-term use. Gait laboratory analysis was done throughout the design process to protect against possible deleterious effects of the new ankle device on gait patterns and the residual limb.

Patent protection of the "entrapment" design was requested. Many standard machine elements were eliminated in the SEATTLE Ankle, such as pivot pins, bearing

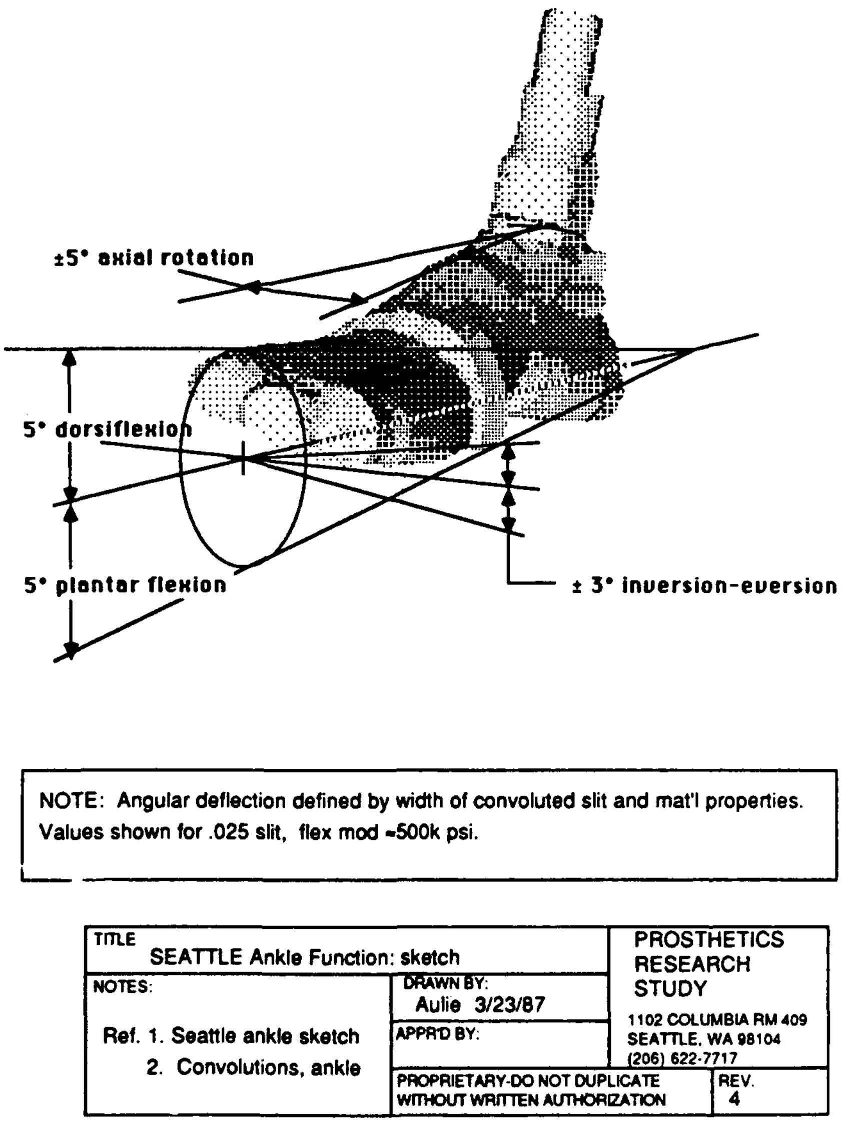

Figure 2.
SEATTLE Ankle function sketch.

surfaces, fasteners, and their associated stress concentrations. The simplicity of this design and its ease of manufacture should result in low start-up costs and in being readily commercially available.

DESCRIPTION

Function

The VA/SEATTLE Ankle (**Figure 1**) is a lightweight monolithic design weighing approximately 300 g (10.5 oz.) with a maximum pylon length of 15 1/2 in. (Actual weight depends on final length of pylon custom-cut per patient.) Energy storage and release and three-axis motion (**Figure 2**) are accomplished via a specifically designed convolution (**Figure 3**) in a distal region (ankle) of the pylon. The convolution in the ankle provides rotation ($\pm$ 5 degrees), inversion/eversion ($\pm$ 3 degrees), and dorsi/plantar flexion ($\pm$ 5 degrees). The flexion/extension of the SEATTLE Foot (and other functionally similar feet) is enhanced when used in conjunction with this ankle.

The ankle functions as follows: 5 degrees plantar flexion allowed at heel strike; $\pm$ 5 degrees elastically resisted axial rotation allowed until mid-stance; dorsiflexion blocking at 5 degrees flexion; and, $\pm$ 3 degrees inversion/eversion resisted at a rate of 15 ft. per pound. This limited

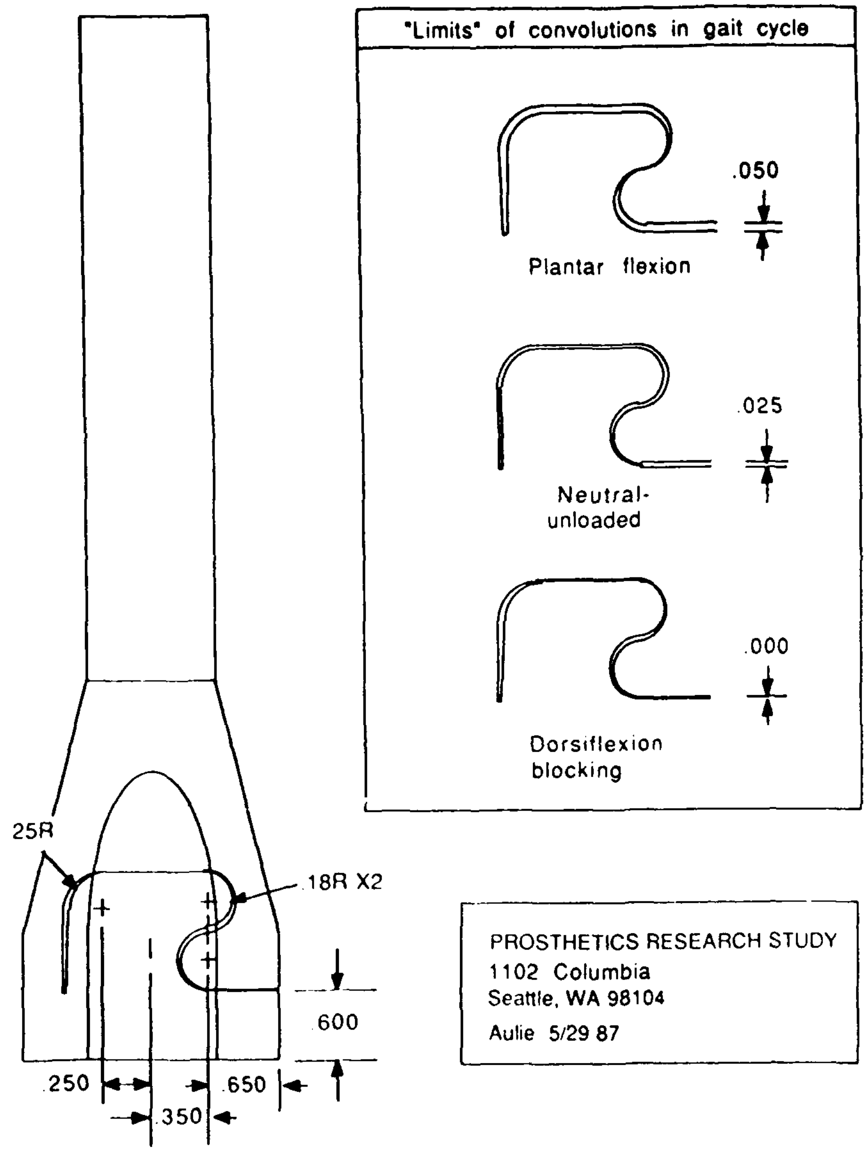

Figure 3.
"Limits" of convolutions in gait analysis.

action provides a range of movement relevant to daily use while providing excellent stability—a very important feature to the geriatric amputee. Note that very rigid dorsiflexion blocking is necessary to operate the SEATTLE Foot and other similar designs.

Installation

Figure 4 illustrates the fitting of the VA/SEATTLE Ankle. The prosthetist was instructed not to sand, cut, glue, or paint the ankle but that marking with pencil or felt-tip pen was acceptable. The following installation features were evaluated:

1. Compatibility with Other Components

Pylons. The VA/SEATTLE Ankle can be interchanged as a substitute for 30 mm pylons from Otto Bock and Teh Lin. Other pylon diameters from USMC, Hanger, etc., which have adapters for 30 mm pylons, can also be used.

Feet. The ankle can be used with any type of SACH Foot with a single bolt attachment. Examples are SEATTLE Foot, Carbon Copy Foot, STEN Foot, SAFE Foot, and all standard and light SACH Feet. It should not be used with Symes, Single-Axis, or Multi-Axis Feet.

2. Length of Pylon

The VA/SEATTLE Ankle should be cut to a length to directly replace the existing endoskeletal SACH Foot

FIGURE 4

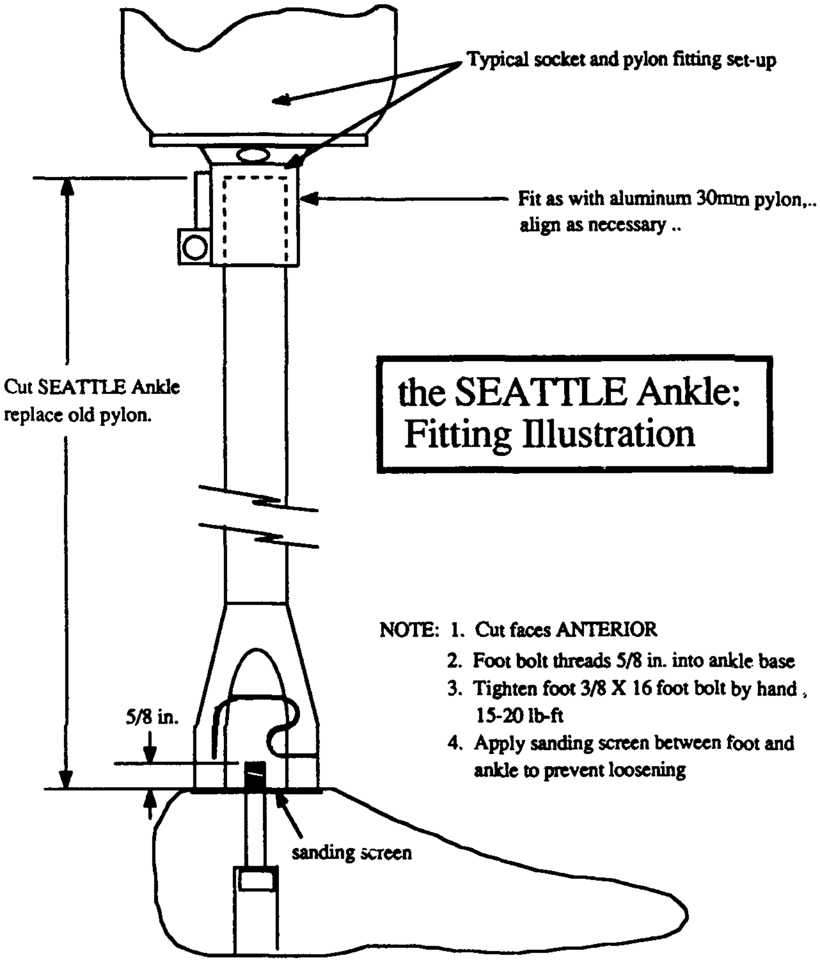

Figure 4.
The SEATTLE Ankle: fitting illustration.

adaptor and pylon. The minimum length required for fitting the ankle is 5 1/2 inches measured from the top of the foot to top of the ankle.

3. Cosmetic Cover

Any existing cosmetic foam cover not need to be modified to fit over the ankle. It may be necessary, however, to enlarge the distal opening of the cosmetic cover to accommodate the distal 2 inches of the ankle.

4. Bolt Applications

The foot is attached to the VA/SEATTLE Ankle according to the foot manufacturer's instructions. The prosthetist was instructed to: use the existing foot bolt only if it is 3/8-16 standard thread (the bolt should protrude a minimum of 1/2 inch and a maximum of 1 inch from above the top of the foot); apply 12 ft. lbs. of torque to the bolt when installing the ankle; and, use the sanding screen supplied as a washer between the foot and the ankle to prevent rotation.

5. Alignment

The VA/SEATTLE Ankle can only be used where the pylon is aligned 90 degrees vertical to the top of the foot at midstance. This applies to both existing and newly fabricated prostheses. For new fabrication, the prosthesis should be aligned first with a temporary prosthetic alignment jig and then transferred to a VA/SEATTLE Ankle in vertical alignment.

CLINICAL EVALUATION METHODOLOGY

General

One hundred and twenty preproduction models of the VA/SEATTLE Ankle were purchased from PRS for a VA nationwide evaluation. Letters from regional directors to 18 VAMC directors initiated the clinical evaluation. This letter was accompanied by a procedures document and

evaluation questionnaires. The SEATTLE Ankles were furnished without cost to the participating VA medical centers by REU.

Subject Selection

Candidates screened could have had either above or below knee temporary limbs. The age of the candidate or activity level was not to be used as a basis for selection. The prime selection criterion was that the candidate had some experience in using a conventional type of pylon without ankle motion.

Evaluation Sites

VA Orthotic Laboratories. Eight VA Orthotic Laboratories were selected for fitting the SEATTLE Ankle on the temporary limbs they normally fabricate. Each facility received six units. The facilities were VAMCs in Decatur, GA, Dallas, TX, Hines, IL, Long Beach, CA, Memphis, TN, Minneapolis, MN, San Antonio, TX, and West Los Angeles, CA.

Commercial Facilities. Ten VA facilities were selected to conduct the clinical evaluation utilizing commercial prosthetic facilities currently having VA contracts. Each facility received five ankles. The facilities were VAMCs in Bay Pines, FL, Boston, MA, Chicago, IL (Westside), Martinez, CA, Richmond, VA, St. Louis, MO, Salt Lake City, UT, Tampa, FL, Washington, DC, and Winston Salem, NC.

Prosthetist Participation

A "Class I Clinical Evaluation Fee" for 1 hour labor at the current contracted rate was authorized in addition to the contracted cost of the endoskeletal limb being ordered. No other additions or deletions were authorized. If it was decided to fit the SEATTLE Ankle to an existing endoskeletal limb, the transaction was considered as a repair. In addition to the "Class I Clinical Evaluation Fee" of one hour labor at the contracted rate, one hour labor also was authorized to cover the prosthetist's time in removing the existing pylon and replacing it with a SEATTLE Ankle.

Clinical Reports

Reporting requirements involved an initial fitting report by the prosthetist and a follow-up report by the prosthetist and subject after the device had been worn for 30 days (see **Appendices**). Longer term reports were desirable but were voluntary. However, it was important to report any failures no matter when they occurred.

Laboratory Tests

The REU determined that cyclic laboratory tests to document durability and reliability over long-term use was necessary. The VA Prosthetics Assessment and Information Center (PAIC), Baltimore, MD, accomplished cycling tests on the ankle connected to a conventional SACH Foot under load. The cycle consisted of heel compression, mid-stance, and toe-break under an average load of 123 lbs.

RESULTS

Laboratory Testing

The laboratory tests conducted by PRS on the development models (11 design iterations and 60 prototypes) indicated a load-bearing capacity up to 2.5 × body weight using a 200 lb. load as standard.

The PAIC's test procedures consisted of 500,000 cycles of the ankle attached to a conventional type SACH Foot under a dynamic load approximating the weight of a 150-170 lb. person. The ankle showed no signs of fracture or wear throughout the entire test procedure.

Clinical Findings

Fifty-three subjects were fitted through the Prosthetic and Sensory Aids Services at 13 VA Medical Centers. REU collected and analyzed the data resulting in the information below. Although 100 ankles were delivered to VAMCs for testing, it was decided to conclude the evaluation at the end of 53 reported cases. This decision was prompted by the extremely small variation—standard error—of reported data. REU continued to receive data on the remaining ankles with the expectation that the data variability would remain constant.

Distribution of Subject(s) Per Field Station

The number of subject fittings does not reflect the original distribution of five or six ankles per VAMC (see **Table 1**).The number reflects the availability of subjects. Ankles were moved from center to center as subjects became available.

Subjects included in the evaluation represented a broad range in age, weight, and activity level. The imbalance in sex reflects the normal veteran population. The heaviest subject (243 lbs.) did not report his activity level. **Tables 2-7** provide breakdowns on subject characteristics.

Subject Response After Using the Ankle for a Minimum of 30 Days

The most active subjects averaged 51 years in age and

Table 1. Geographical distribution

VAMC	Subjects fitted
Chicago, IL (Westside)	3
Dallas, TX	5
Decatur, GA	4
Hines, IL	3
Los Angeles, CA (West)	11
Martinez, CA	6
Memphis, TN	6
Minneapolis, MN	2
Richmond, VA	1
St. Louis, MO	1
Salt Lake City, UT	5
San Juan, PR	5
Winston-Salem, NC	1
Total	53

Table 2.

Sex	
Male	92%
Female	8%
Total	100%

Table 3.

Age distribution	
23-30	4%
31-40	19%
41-50	17%
51-60	17%
61-69	19%
70+	11%
Unknown	13%
Total	100%

Table 4.

Weight range	
0-150 (lb)	17%
151-199	62%
200+	19%
Unknown	2%
(heaviest = 243/lbs.)	
Total	100%

ranged in weight from 170 to 227 lbs. They were likely to wear an energy storing/release foot and considered the SEATTLE Ankle an asset. This group in general would accept increased performance of the ankle motion, especially eversion and rotation.

The medium activity group averaged 49 years in age and ranged in weight from 131 to 230 lbs. They had less than 50 percent chance being a wearer of an energy storing/release foot, but also considered the SEATTLE Ankle an asset. The anomaly concerning a slightly lower average age group having medium activity was not considered significant. The salient feature was the use of an energy storing/release foot.

The lowest activity group were older, averaging 63 years and they ranged in weight from 140 to 235 lbs. They did not use an energy storing/release foot but considered the ankle adequate and generally helpful to them.

Table 7 shows the assessment of the three groups of subjects. **Appendix C** represents the form used by subjects in evaluating the ankle after 30 days of use.

The subjects' responses when asked to rate their performance during certain activities with the SEATTLE Ankle as compared to what they previously used is summarized in **Table 8**. The average response was calculated

Table 5.

General activity level*	
High	36%
Medium	38%
Low	25%
Unknown	1%
Total	100%

***General activity level** is an indicator on how much and what type of activities the subject does.
High: frequent walking and ambulatory activities; active in (ambulatory) sports.
Medium: moderate amount of walking and ambulatory activities; some (ambulatory) sports.
Low: minimal amount of walking and ambulatory activities; no (ambulatory) sports.

Table 6.

Amputation level	
Below-knee	87%
Above-knee	11%
Unknown	2%
Total	100%

for the subjects as a whole and then by activity level groups. Improved performance in all ambulatory functions is explicitly shown in the ratings given by the subjects for walking either slow or fast (7.5 for slow, 7.3 for fast), running (6.0), going up stairs (7.3) or down stairs (7.4), going up hills (7.1) or down hills (6.9), and going over uneven terrain (7.5). The group of subjects with the highest activity level showed an even greater improvement rating. **Table 8** shows that: (1) moderate improvement was achieved in running; (2) high activity persons detected downhill improvement; and, (3) high activity subjects consistently improved.

Table 9 presents the subjects' rating of the degree of rotation provided by the ankle.

Prosthetist Interpretation

The prosthetists reported that fittings went comparatively easily except in six instances were the pylon diameter did not directly fit the alignment jig or knee receptor. The form used by the prosthetists for fitting evaluation is presented in **Appendix A**. Prosthetists' comments and interpretation of subjects' feedback after a minimum of 30 days use by the subject is shown in **Table 10**. Data are summarized by percentages within groupings of the subjects' response when questioned on degree of specific ankle function. A prosthetist form for 30-day follow-up subject evaluation is presented in **Appendix B**.

Ninety-eight percent of the subjects reported that ankle movement was smooth during all phases of gait. The functioning of the ankle was essentially noiseless and failure free. Only three subjects reported functional failures during the evaluation period. These failures were caused by: (1) a bolt broken while playing golf; (2) a helicoil pulled out during running; and, (3) a bolt broken while walking on uneven terrain.

Comments by Subjects and Prosthetists

A large percentage of subjects indicated that the defined motion of the SEATTLE Ankle as tested was acceptable. Results on two functions require addressing: (1) the degrees of plantar flexion, and (2) rotation. The subjects in the highest activity group preferred more plantar flexion, while those in the medium and low activity groups desired it to stay as was. Secondly, 53 percent of the subjects wanted to keep the amount of rotation as it was, while 23 percent wanted more rotation. In opposition to this result is the

Table 7. Qualitative assessment of the ankle by activity level, age, and type of foot

General activity level	No. of subjects	Average age	Using SEATTLE Ankle an improvement	Used special foot
High	19	51	73%	63%
Medium	20	49	65%	30%
Low	13	63	38%	0%

NOTE: Special foot means one that stores and releases energy during use (e.g., the SEATTLE Foot).

Table 8. Performance level by activity function

General activity level	Walk slow	Walk fast	Run	Go upst.	Go dwnst.	Up hill	Down hill	Uneven terrain
All subjects	7.5	7.3	6.0	7.3	7.4	7.1	6.9	7.5
High	7.9	6.3	6.5	7.7	7.8	7.9	8.0	7.6
Medium	7.8	7.6	6.1	7.3	7.3	7.0	6.2	7.4
Low	7.0	6.8	5.6	6.8	6.9	6.3	6.4	6.8

Rating Scale: 1 (worse than) . . 5.5 (equal to) . . 10 (easier than)

prosthetist interpretation of the subjects' requirements, which indicated that 55 percent wanted more, while 45 percent wanted to keep it as was.

The comments from the prosthetists to broaden the compatibility of the SEATTLE Ankle to mate well with existing alignment jigs and prosthetic knee units were reviewed by the PRS development team and the prospective manufacturer. Prosthetists' suggestions for improvements were primarily concerned with mating the ankle's pylon to a broader variety of knee receptors and below-knee alignment jigs (e.g., the Berkeley Alignment Jib and the Hosmer Ultra Roelite Modular Knee).

CONCLUSION

The VA/SEATTLE Ankle unit seemed to have met the design goal of being a replacement for most below-knee and above-knee endoskeletal pylons. The experiences of persons with lower limb amputations who used the SEATTLE Ankle throughout this evaluation endorsed the success of the project goal for improving gait. Selected comments by subject users follow.

"Yes, I'm walking better and have more spring action using the SEATTLE Ankle."

"I prefer the SEATTLE Ankle, it allows easier walking and is more flexible than my previous prosthesis."

"Yes, the SEATTLE ankle is much lighter and restores some lost ankle movements."

Common descriptor used by subjects evaluating the VA/SEATTLE Ankle include *"more natural gait," "better traction," "flexibility," "lightweight," "improved gait," "more stability," "more bounce or spring to my step," "feels more natural and is easier to walk with."*

This evaluation determined that the performance of the VA/SEATTLE Ankle proved to be acceptable and reliable with a wide variety of users and its commercial production was recommended.

Table 9. Percentage* of subjects reporting on ankle rotation

General activity level	Just right	Too stiff	Too loose	Unknown
All Subjects	53.0	23.0	0	24.0
High activity	20.0	9.5	0	—
Medium activity	15.0	9.5	0	—
Low activity	18.5	5.0	0	—

*Percentage of respondents within each group.

Table 10. Prosthetist interpretation of Ss desire for ankle function in percent

General activity level	Plantar flexion More	Plantar flexion Same	Dorsi flexion More	Dorsi flexion Same	Eversion More	Eversion Same	Inversion More	Inversion Same	Rotation More	Rotation Same
All subjects	45%	52%	45%	52%	51%	49%	49%	51%	55%	45%
High activity	19	9	16	18	21	15	21	14	25	12
Medium activity	16	25	13	24	15	20	16	20	22	15
Low activity	10	18	16	10	15	14	12	17	8	18

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APPENDIX A
Prosthetist's Evaluation: Fitting

TO: Director, VA Rehab R&D Evaluation Unit
Prosthetics R&D Center
103 So. Gay Street
Baltimore, MD 21202

FROM: CHIEF, PROSTHETIC SERVICE, VAMC ______

SUBJ: Prosthetist's Evaluation — Seattle Ankle Fitting

Veteran's Name ______ Date ______
Birthdate ______ Sex ______ SC ______ NSC ______
Amputation Side ______ Level ______ Length of Stump Overall ______
Weight ______ Height ______
Activities ______
______
Activity Level (Low, Medium, High) ______
Prosthesis Description ______
______
______ Weight ______
Foot Type ______ Foot Size ______ Shoe Size ______

Part Number ______
Cut Length Overall ______ Weight (cut) ______
3/8" Bolt Length ______ Wt. of Pylon Removed ______

Prosthetist ______
Prosthetist's comments on fitting ______
______
______
______
______

Prosthetist's comments on function: (plantar-dorsiflexion, inversion, eversion, rotation) ______
______
______
______
______
______
______

Prosthetist's interpretation of Subject impressions of function ______
______
______
______
______
______
______

Notes* *Each ankle has undergone bench testing to assure its safety.* Should failure occur for any reason, call the Prosthetics Research Study (206) 622-7717 and return the part to VA Rehab R&D Evaluation Unit, Prosthetic R&D Center, 103 South Gay Street, Baltimore, MD 21202

APPENDIX B
Prosthetist's Evaluation: 30 Day Follow-Up

TO: Director, VA Rehab R&D Evaluation Unit
Prosthetics R&D Center
103 South Gay Street
Baltimore, MD 21202

FROM: Chief, Prosthetic Service, VAMC ______

SUBJ: Prosthetist's Evaluation — Seattle Ankle 30-day Follow-up

Date ______ Part ______
Veteran's Name ______ Weight ______
Part Number ______
Prosthetist ______

Prosthetist's interpretation of Subject feedback ______

Would Subject like "more"/"less": Rotation . . . .
Inversion . . . .
Eversion . . . .
Plantarflexion . . . .
Dorsiflexion . . . .

Is Ankle movement smooth during all phases of gait? ______
If no, explain ______
Has Ankle movement increased since fitting? ______
In what ways? ______

Noise? ______ When? ______
From where? ______
Failure? ______ When? ______
Where? ______ What was Subject doing? ______

Did the foot loosen from the Ankle? ______
What was used to affix the foot to the Ankle? (hot glue, sanding screen . . .) ______
Prosthetist's suggestions for improving Ankle design ______

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APPENDIX C
Subject's Evaluation: 30-Day Follow-Up

TO: Director, VA Rehab R&D Evaluation Unit
Prosthetics R&D Center
103 South Gay Street
Baltimore, MD 21202

FROM: Chief, Prosthetic Service, VAMC ______________________________

SUBJ: Veteran Questionnaire — SEATTLE ANKLE — 30-day Follow-up

Veteran's Name ______________________________ Date ______________
Part ______________________________
Have you used the Ankle on a daily basis? ______________________________
Have you used the Ankle while participating in sports? ______________________________
During what specific activities have you used this Ankle? ______________________________
____
____
____

Please rate the difficulty to perform the following activities using your new Ankle as compared to the ankle that you wore previously . . .

	Worse 1	2	3	4	Same 5	6	7	8	Easier 9	10
Walk slowly										
Walk fast										
Run										
Go up hill										
Go downhill										
Go upstairs										
Go downstairs										
Uneven terrain										

Are there any other specific activities where the function of the Ankle has been noticeable? Which? ______________
____

How would you rate the degree of rotation (twist) provided by the Ankle . . . [ ] Just Right [ ] Too Stiff [ ] Too Loose
Do you have any other comments, suggestions, or criticisms with regard to this Ankle design that might help us improve it for other amputees? ______________________________
____
____

Do you consider using the Ankle an overall improvement over using your prosthesis without the Ankle? ______________
Why? ______________________________
____

Department of Veterans Affairs

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CLINICAL REPORT

An Evaluation of Capuchin Monkeys Trained to Help Severely Disabled Individuals

Prepared by the Rehabilitation R&D Evaluation Unit, Rehabilitation Research and Development Service, Department of Veterans Affairs, Baltimore, MD 21202

Abstract—This report describes an evaluation made by the Rehabilitation R&D Evaluation Unit (REU) of research conducted jointly by Helping Hands, Inc., Boston, MA and Boston University which was funded by the Rehabilitation Research and Development Service, Department of Veterans Affairs. The report covers an assessment conducted in February and March 1989 to determine the activities, effectiveness, advantages, and disadvantages exhibited by the capuchin monkeys placed as aides in the residences of disabled persons.

Key words: *Capuchin monkey aides, evaluation, quadriplegics, severely disabled.*

INTRODUCTION

The results of 10 years of research and development activities concerning the use of monkey helpers to quadriplegics were evaluated by assessing current placements. Helping Hands, Inc. reported that a total of 14 monkey helpers have been placed.

In this evaluation, a detailed retrospective study was made of nine of the ten monkeys currently placed as helpers to quadriplegic individuals. One placement was considered geographically too distant to be included in the study. An experienced therapist conducted interviews with the owners of eight of the monkeys who were living at home. The monkeys were directly observed performing their assistive tasks. One owner was interviewed in the hospital.

For further information, contact Saleem J. Sheredos, acting director, Rehabilitation R&D Evaluation Unit, VA-PRDC, 103 S. Gay St., Baltimore, MD 21202. Phone: 301-962-2133

In the four placements no longer active, telephone interviews were conducted with the disabled persons in two cases and an on site visit was conducted with the wife of the disabled person now deceased in one case.

Helping Hands Inc. was visited by the evaluators as well as the monkey breeding colony in Disney World. Additionally, the interviewers visited the homes of three of the 89 foster parents reported by Helping Hands as raising monkeys to be trained as helpers.

The evaluation provided sufficient positive findings that monkey helpers are useful, effective and acceptable.

METHODOLOGY

Helping Hands, Inc. identified 14 placements of monkey helpers. This report summarizes data from interviews conducted with nine disabled persons who received monkey helpers.

Eight of the nine active monkey placements were observed directly by an experienced therapist. The recipients of the monkey helpers, all of whom were quadriplegics, were interviewed by the therapist. One person was hospitalized at the time and the interview was conducted in the hospital. The interviewer spent at least 3 and one-half hours at each of the interview sites.

The circumstances of the remaining five placements are summarized below:

1. A visit was conducted with the spouse of a deceased veteran who had received a monkey helper. The monkey, who was awaiting reassignment, was still present in the home during the interview. The wife reported that she and

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her husband were pleased with the monkey helper and that it had been possible for him to be left alone with the monkey for 5 to 6 hours. Previously, she could not leave her husband unattended. The interviewer observed that the monkey responded to her verbal commands.

2. A visit was conducted with a quadriplegic recipient and his father. At the time of the interview the monkey had been returned to Helping Hands, Inc. for reassignment. The recipient and his father reported that for 5 months the monkey had been very helpful. Then the recipient was bedridden and isolated for 3 months, after which the monkey responded only to the father.

3. Telephone contact was made with another person who had received a monkey helper but subsequently returned it to Helping Hands. (This monkey helper was reassigned to one of the other recipients who was site-visited in this study.) The initial owner reported that for 1 year the monkey performed when requested and as needed. However, when the owner returned to work the monkey was left alone for many hours during the day and it was felt that the monkey would better serve someone else.

4. One placement was made overseas, which was considered beyond the travel range for this evaluation.

5. One placement was discontinued because the recipient, who suffered from cerebral palsy, lacked sufficient motor control to properly operate the monkey-related equipment. This monkey helper was returned to Helping Hands for reassignment.

FINDINGS

A. Environment of the Monkey Helper

The monkey helpers in this evaluation were placed with six male and three female quadriplegic individuals. One recipient's disability resulted from muscular dystrophy, the remaining eight were due to traumatic spinal cord injury. Two recipients, both male, were veterans.

Six of the disabled persons resided with a family member (four of these with a spouse), two in a resident home, and one lived alone with a 24-hour attendant. Three of the recipients had a part-time attendant. Five of the recipients had pets (cats, dogs, and a ferret), none of which seemed to pose problems for the monkey helper.

The family members universally accepted the presence of the monkey helper. Seven caregivers reported free time on a daily basis as a result of a few hours of assistance by the monkey helper. All nine of the disabled persons reported that the monkey helper was a pet and companion, as well as an aide.

Table 1
Number of Assistive Tasks Performed by a Monkey Helper (Reported to and Observed by Interviewer)

Name of monkey	Number of years with disabled person	Number of tasks
Hellion	10.0	28
Freeway	8.0	24
Su Su	6.5	28
Henri	6.0	28
Jo	5.0	27
Peepers	1.5	23
Cleo	0.5	15
Jeep (reassignment)	0.5	6
Maggie	0.4	23

B. Description of Monkey Helpers

The monkey helpers in this study were placed for varying periods of time (see **Table 1**). Three of the monkeys (Jeep, Cleo, and Maggie) were placed for less than 1 year, whereas, Hellion was placed for about 10 years.

Capuchin monkeys live an average of 30 years. The monkey helpers reported on ranged in age from 6 to 20 years old. The productive life expectancy of the Capuchin is estimated to be 20 years.

C. Activities Performed

All monkey helpers are trained in a basic repertoire of 60 assistive tasks (see **Appendix A**). Although a monkey helper arrives at the home of the disabled person with this training, it is the particular needs of the individual that will determine what specific activities the monkey will ultimately perform.

In this study, the monkey helpers were observed to perform about 58 percent of the tasks which they learned at Helping Hands, Inc. Additionally, they were able to perform other tasks not addressed in basic training (e.g., turning on or off a computer, television, radio and lights; repositioning a foot on a footrest; pushing buttons on a telephone; and turning a faucet on and off).

The types of tasks requested of the monkey helper varied according to the individual's needs. All individuals reported complete satisfaction with the monkey helper's response, performance, and behavior.

Each person interviewed stated that the monkey accomplished all of the requested tasks by responding to verbal commands; the laser pointer was used in a demonstration only, and there was no need for negative shock reinforcement. The shock pack, a device built into a belt

which the monkey wears and which allows the trainer to sound a warning tone or give the monkey a 0.5 second shock, was reported as being used only during the initial in-home orientation or in training of new tasks.

A review of the tasks performed by the monkey helper indicated that the longer the monkey lived with the disabled person, the greater the number of tasks it performed (see **Table 1**).

The assistive tasks observed by the interviewer as they were being performed by a monkey helper are listed in **Appendix B**. These tasks can be sorted into four categories: feeding, fetching, manipulating objects, and personal care. The monkey helpers all responded very reliably to verbal commands and to the laser pointer in a demonstration. The monkeys did what they were asked to do by their owner.

The survey results showed that the monkey helper increased the owner's independence of human assistance. This is explicitly reflected by the increased time that the recipients spent without a caregiver since receiving a monkey helper (see **Table 2**).

Flexibility of the monkey as a helper is important. The monkey should be able to accompany its owner to perform requested tasks anywhere in the house and even travel with the disabled person. The demonstrated ability to learn additional tasks not taught in basic training (e.g., foot positioning) expands the utility of these helpers.

A very important aspect clearly experienced by all persons interviewed was the relationship that developed between the recipient owner and the monkey helper. In all cases this "bonding" meant a lot to the owner, who viewed the monkey as a companion and pet in addition to functioning as an aide.

D. Acceptability

A number of areas were reviewed regarding the acceptability of a monkey helper. The following potential problem areas were explored by the interviewer but produced no unfavorable information.

- *Aggressive or undesirable monkey helper behavior.* All interviewees identified their monkey's behavior as acceptable and none reported incidents of unacceptable behavior.
- *Care of monkey.* The feeding, caging, grooming, etc. of the monkey was acceptable to the family, and/or attendant.
- *Health care of monkey.* The required medical care of all monkey placements was reported as minimum to nonexistent. Two instances of colds were reported which, in coordination with Helping Hands, were treated locally in a routine manner. In view of the length of time (5 to 10 years) the monkeys have lived with the disabled person, there appear to be no special medical or health care requirements. The eight monkeys visited appeared healthy and clean. No odors were detected in the house.

Table 2
Hours a Quadriplegic Spends Without a Caregiver

	Without monkey helper	With a monkey helper
Average hours	0.4 hours	5.9 hours
Range for 9 quads	0 to 3 hours	1 to 15 hours

- *Matching of monkey with disabled individual.* The screening and preplacement work-up established by Helping Hands, Inc. appeared to be effective. At the time of interviews, monkeys were functioning well. In the terminated placements, the monkeys were removed to respond to the changing needs of the recipient.
- *Supply of monkey helpers.* Helping Hands, Inc. has developed a system to breed, socialize, and train monkeys as helpers. The span of 4 to 5 years from birth to completion of training of each monkey affects the supply of monkey helpers on an annual basis. Helping Hands, Inc. reports 89 monkeys are currently in foster homes as part of the preplacement process. Fifteen monkeys were in training at Helping Hands, Inc. at the time of this evaluation.
- *Care of monkey helper in absence of disabled person.* Absence of the disabled owner, either for short-term hospitalization or employment, presents problems. The monkey helper tends to regress in terms of learned behavior and may transfer allegiance or response patterns to someone else in the household. Helping Hands, as a part of its placement criteria, makes a systematic determination on the disposition of the monkey helper should absence of the owner occur.

CONCLUSIONS

Monkeys can be taught to perform a variety of useful tasks for disabled individuals. By performing these tasks, the monkey helper offers a degree of freedom for a caregiver. A monkey helper also can provide the opportunity for the disabled person to be without a caregiver for periods of time. The performance of assistive tasks is a significant positive contribution that enables a disabled individual to become more independent. Sufficient posi-

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tive results were obtained in this evaluation to conclude that monkey helpers are useful and accepted.

The monkey helper was positively accepted in the household of each of the nine disabled individuals and by all types of caregivers, friends, family, and paid attendants. Monkey helpers exhibited a positive behavior and became accepted members of the household.

The research, development, and evaluation phases of this project are completed. The evaluation of the monkeys placed was found to be satisfactory. If enabling legislation is passed authorizing the Department of Veterans Affairs to provide monkey helpers to quadriplegic veterans, Helping Hands, Inc. would be considered a vendor providing the monkey helpers and all necessary related services including placement and follow-up as required.

Appendix C presents a discussion of the results of a survey to determine veteran interest in concept of monkeys as helpers for disabled individuals.

APPENDIX A
Basic Repertoire of Tasks Monkey Helpers Are Trained to Perform by Helping Hands, Inc.

Feeding

Hold sandwich
Spoon-feed meals
Feed snack

Personal Care

Wipe face
Scratch face with cloth
Put on eye glasses
Take off eye glasses

Fetching

Retrieve mouth stick
Retrieve bottle
Retrieve cup
Retrieve plate
Lift small items from floor
Lift small items from table
Place book for reading
Retrieve splint

Manipulating Objects

Reposition quad's arm
Open cage door
Close cage door
Load floppy disk
Pull cords
Position remote controls
Turn book pages
Position computer paper

Manipulating Objects

Open bottle cap
Close bottle cap
Place drinking straw
Remove drinking straw
Open small refrigerator door
Close small refrigerator door
Place sandwich in holder
Hold cup
Hold plate
Hold bottle
Open oven door (cold)
Close oven door (cold)
Open food container
Close food container
Open cartons
Close cartons
Vacuum floor
Open doors
Close doors
Lock doors
Unlock doors
Open drawers
Close drawers
Open cabinet
Close cabinet
Empty trash
Clean table top spills
Open curtains
Position "J" stick
Load VCR cassette
Unload VCR cassette
Load audio cassette
Unload audio cassette
Position sip and puff unit
Position small appliances
Change floppy disk
Position disk for CD player

APPENDIX B
Assistive Tasks Observed in this Study as Being Performed by a Monkey Helper

Feeding

Spoon-feed meals
Feed snacks
Hold sandwich

Personal Care

Wipe quad's face
Scratch quad's face

Fetching

Retrieve bottle
Retrieve cup
Retrieve plate
Retrieve mouth stick
Lift small items from floor
Lift small items from table
Place books for reading

Manipulating Objects

Wipe table top
*Wipe lap tray
Place straw
Remove straw
Place sandwich holder
Remove bottle top
Replace bottle top
Hold Bottle
Hold cup
Position "J" stick
Open food container
Close food container
Open small refrigerator
Close small refrigerator
*Turn on computer
*Turn off computer
Open cartons
Close cartons
*Turn on lights
*Turn-off lights
Lock door
Unlock door
*Turn-on radio
*Turn-off radio
Load audio cassette
Unload audio cassette
Load video cassette
Unload video cassette
Position quad's arms
*Position foot on footrest
Position sip and puff unit
*Turn-on faucet
*Turn-off faucet
*Push buttons on telephone

*Not included in basic repertoire training

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APPENDIX C
Summary Report of Survey to Determine Disabled Veterans' Interest in Capuchin Monkeys as Aides

To determine the potential need and desirability of monkey helpers as care providers to quadriplegic veterans, a survey was conducted in March, 1989. The survey was developed through a joint effort by the VA Rehabilitation Research and Development Service, M.J. Willard, Ed.D., principal investigator, and Boston University, and was mailed to 1,000 quadriplegic veterans (500 service-connected injury and 500 nonservice-connected injury). The results were tabulated and analyzed by an independent consultant.

The survey was completed and returned by 151 veterans (both service- and nonservice-connected injuries). Among the respondents, the average level of spinal injury was C-6, with ranges from C-1 to T-1. The average time spent in a wheelchair was 6 to 8 hours a day. Eighty percent of respondents lived at home with a spouse, 15 percent resided in a nursing home, and 5 percent were in a hospital (not determined whether intermediate medicine or spinal cord injury long-term care). Sixty-five percent had children in the residence and 55 percent had attendants. Approximately 4 percent stated that they worked.

An essential question of the survey questionnaire was: "Considering both positive and negative traits of monkeys, would you want a monkey-helper? There were 29 "yes," 102 "no," and 20 "maybe" responses.

In the "yes" category, there were 26 male respondents and 3 female respondents. However, not all the "yes" answers were unequivocal; many had qualifying statements relating to conditional or trial acceptance and need for more information. None in the "yes" category responded that they worked.

In the "maybe" category there were conditional and/or provisional statements that usually reflected or stated "not now, but maybe later."

Respondents in the "no" category, as well as some in the "maybe" category, found several areas of monkey helper behavior and traits not acceptable. The following percentages reflect the frequency of indication of an unacceptable trait:

Objections of family member(s) — 85%
Objections to general animal traits/behavior — 70%
Objections to a dependent animal — 65%
Objections from an attendant — 50%
Objections to animal sexuality — 45%
Away from home too much — 30%
Cost of care and feeding — 5%
Would interfere with work — 3%
Other objections — 25%

Conclusions

A review of the returned questionnaires showed that about two-thirds of the respondents either did not complete all of the information requested or did not adequately and/or accurately fill in the necessary information. Responses did indicate that many of the veterans considered owning a capuchin monkey to be in the "pet" area. Some respondents had personal reservations as to the length of stay of a monkey. Objections from family members or an attendant were listed as reasons, as well as cost of maintenance, and undesirability of a dependent animal, animal traits, and behaviors.

Some respondents viewed a monkey helper as a temporary visitor who could be returned at any time. As the risks of worsening physical or psychosocial conditions increase, concern for the monkey could become an issue, as it would when children, family "live-ins," and additional pets were added to the monkey's environment.

Serious consideration must be given to the fact that 1,000 survey forms were sent out and only 151 were returned, which would infer a lack of interest or ambivalence. The feasibility/needs analysis indicated that interest is not great in the total number of respondents. However, those who were interested were positive in their approach and realistic in expectation as to the role the monkey would play in care delivery.

ABSTRACTS OF RECENT LITERATURE

by
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and
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Professor Emeritus of Sensory Rehabilitation, New York University, and David Peikoff Chair in Deafness Studies, University of Alberta, Edmonton, Canada

Abstracts are drawn primarily from the orthotics, prosthetics, and sensory aids literature. Selections of articles were made from these journals:

Assistive Technology
Canadian Journal of Rehabilitation
Journal of Bone and Joint Surgery
Journal of Medical Engineering and Technology
Journal of Speech and Hearing Disorders
Journal of Speech and Hearing Research
Journal of Visual Impairment and Blindness
Prosthetics and Orthotics International

PROSTHETICS, ORTHOTICS, AND RELATED TOPICS

Accelerations Due to Impact at Heel Strike Using Below-Knee Prosthesis. Van Jaarsveld HWL, Grootenboer HJ, De Vries J, reprinted from *Prosthet Orthot Int* 14:63-66, 1990.

The acceleration in the sagittal plane of the prosthetic tube at heel strike in normal walking was measured in five health amputees with their definitive below-knee prosthesis, every subject using six different prosthetic feet, wearing sport shoes as well as leather shoes. The experiments were carried out in the rehabilitation centre "Het Roessingh," Enschede, The Netherlands.

Maximum accelerations were extracted from the acceleration-time-signal. Mean acceleration maxima of all subjects were calculated for each foot-shoe combination to eliminate the individual influence of the subjects. In the axial direction the maximal accelerations demonstrate a clear difference among the prosthetic feet and the shoes, while in dorsoventral (tangential) direction the inter-individual variation in the acceleration extremes dominates the difference between the types of footwear. In comparison with nonamputees the magnitude of the maximal axial acceleration at heel strike does not differ significantly. [JEE]

Commercial Software as the Basis for an Augmentative Communication System on a Personal Computer. Horstmann HM, Levine SP, Kett RL, reprinted from *Assist Technol* 2:19-26, 1990.

Many severely speech-impaired individuals have a need or desire to maintain the capability for computerized speech generation while performing a variety of other computer-supported tasks such as word processing, financial analysis, or database management. Criteria are presented for augmentative communication systems that both incorporate a personal computer (PC) and provide capabilities for voice output to satisfy this requirement. Current approaches to such systems are categorized and reviewed in light of these criteria. A new method is then described in which word processing or other appropriate PC software is used for speech production, providing a natural integration of written and spoken communication. The approach separates the computer access and voice output functions to allow great flexibility in the choice of an alternative access system. This flexibility combined with appropriate exploitation of PC software features has the potential to yield a high communication rate as well as maximum compatibility with application software. [JEE]

A Critical Review of Laser Doppler Flowmetry. Obeid AN, Barnett NJ, Dougherty G, Ward G, reprinted from *J Med Eng Technol* 14:178-181, 1990.

Laser Doppler flowmetry (LDF) is now a well-established, noninvasive technique for measuring microvascular blood perfusion. However, there are a number of factors which can seriously affect the interpretation of the laser Doppler signal which are often not considered during routine use. These include: consideration of signal processing limitations, choice of processing bandwidth, problems with motion artefact and instrument calibration, the effect of probe pressure on the skin, and the type of laser used. This paper reviews many of the problems and limitations frequently encountered using the laser Doppler technique. [JEE]

Effect of Weight-Bearing on Healing of Cortical Defects in the Canine Tibia. Meadows TH, Bront JT, Chao EYS, Kelley PJ, reprinted from *J Bone Joint Surg* 72-A:1074-1080, 1990.

It has been generally accepted that mechanical stimulation is an important factor in the promotion of formation of bone. Fracture-healing consists of periosteal bridging of the fracture, which achieves stability, and proliferation of endosteal bone to fill the defects between the ends of the bone. To evaluate the effect of weight- bearing on bone-healing, an operatively created defect in the tibial cortex was chosen as an experimental model. In one set of dogs (Group 1), a bilateral defect in the tibial cortex was created and weight-bearing was permitted on one tibia but not on the opposite one. In Group 2, a bilateral defect in the tibial cortex was made and weight-bearing was allowed on both tibiae. A third group of dogs of similar age (Group 3) had no tibial defects. Quantitative histomorphometry was used to measure formation and porosity of bone. Weight-bearing was measured with both static and dynamic techniques.

Significantly less woven bone formed in the defects in the non-weight-bearing tibiae than in the weight-bearing tibiae. This appeared to be due to a disuse response in the underloaded tibiae, in which less bone formed, rather than to the formation of more bone in the weight-bearing tibiae. The data suggest that weight-bearing is a permissive factor, not a stimulus, for formation of woven bone in a tibial defect.

Clinical Relevance: This animal model supports the concept that lack of weight-bearing decreases the amount of woven bone that is formed in a healing tibial defect. The results of this study indicate that weight-bearing increases the formation of bone fracture-healing. Therefore, it is possible that, in the clinical situation, weight-bearing has a positive effect on bone-healing. [JEE]

An Electromagnetic Cervical Range-of-Motion System and Its Sensitivity to Environmental Metal. Schendel MJ, Patterson R, Allison J, reprinted from *Assist Technol* 2:27-32, 1990.

A three-dimensional cervical spine range-of-motion system was developed using an electromagnetic tracking system for data collection and a personal computer for analysis and graphic representation of the data. A test was designed to investigate the sensitivity of the electromagnetic device to the proximity of metal. It was found that position errors could be kept below 0.5 cm, and rotation errors could be kept below 1.3 degrees, if all metal was at least 33 cm from the source and sensor. The sensitivity of the system to metal should not cause serious problems in the typical clinical environment if simple precautions are taken. [JEE]

Epidemiology of Wheelchair Accidents: Lessons from the Boston Marathon. Kirby RL, reprinted from *Assist Technol* 2:59-68, 1990.

During the 1987 Boston Marathon, 14 of the 46 wheelchair participants tipped over in their chairs during a 13-s chain reaction of accidents near the bottom of the first hill. An analysis of the available documentation revealed factors that may have contributed to the accidents—the tight grouping at the start, the wetness of the road, the steepness of the hill, and speeds that were well in excess of the average for the course. Nine of the 14 tipping accidents were lateral in direction; 5 of the 14 subjects did somersaults in their chairs. Although there were no serious injuries, four of those involved were unable to complete the course, some due to mechanical failure. In 1988 and 1989, a pace car was used at the start, and there were no accidents, although the number of participants and the conditions were similar. This analysis provides insights into how the likelihood of serious injury might be further reduced. [JEE]

Gait Recovery Pattern of Unilateral Lower Limb Amputees During Rehabilitation. Baker PA, Hewison SR, reprinted from *Prosthet Orthot Int* 14:80-84, 1990.

The aim of this study was to determine the rate at which gait recovery as measured by temporal distance factors (velocity and symmetry) occurs in unilateral lower limb

amputees. A microcomputer footswitch system was used to record the gait patterns of twenty subjects, mean age 65.1 years. The initial measurement was taken when the subject was capable of walking 6 meters with an interim prosthesis within the parallel bars. The patient sample as a whole was analyzed and subjects were further divided into four groups, depending on ambulatory aid required at discharge. Group A, n = 3 used no aid, Group B, n = 5 used a single stick, Group C, n = 6 used 2 single sticks and Group D, n = 5 required frames. A one way analysis of variance ($F = 4.55$, $p = 0.02$) showed a significant difference between the Groups, (A and D, B and D, C and D). The major velocity increase occurs within the first 30 days of the gait training program. Overall about 55 percent increase in velocity can be expected within the first fifteen day period followed by an additional 30 percent between days 15-30. A moderately strong correlation ($r = 0.78$) was found between initial and discharge velocity. The correlation between initial and discharge symmetry was weaker ($r = 0.50$). [JEE]

Long Term Comparison of Some Shock Attenuating Insoles. Pratt DJ, reprinted from *Prosthet Orthot Int* 14:59-62, 1990.

The effect of one years general use on the performance of four shock attenuating insoles is reported. Testing was carried out using the JP Biomechanics Shock Meter on twelve volunteers on a timed oval course at eight intervals during the year. The results show that two of insoles perform well (Viscolas and PPT) although deterioration does occur after 6 to 9 months use; the other two insoles (Plastazote and Gait Aid) perform poorly. It is suggested that manufacturers provide some information to the user or supplier regarding the effective life of their products. [JEE]

Lower Limb Sympathectomy Assessed by Laser Doppler Blow Flow and Transcutaneous Oxygen Measurements. Lantsberg L, Goldman M, reprinted from *J Med Eng Technol* 14:182-183, 1990.

In retrospective study of critical ischemia of the lower limb, sympathectomy appeared to be of value in the majority of patients. We therefore assessed sympathectomy by measuring skin blood flow before and after the procedure using laser Doppler flowmetry (LDF) and transcutaneous oxygen tension ($TcPO_2$) techniques. Twenty patients underwent chemical sympathectomy and there was one surgical procedure. Measurements were performed before and 1 week after sympathectomy below the knee and on the forefoot. Symptomatic improvement occurred in 20 of 21 patients. This study demonstrates that skin blood flow in the leg and foot is improved by sympathectomy and confirms objectively our clinical impression. [JEE]

Lumbar Discography in Normal Subjects: A Controlled, Prospective Study. Walsh TR, Weinstein JN, Spratt KF, *et al.*, reprinted from *J Bone Joint Surg* 72-A:1081-1088, 1990.

Major advances in the techniques of discography since 1968, in conjunction with major strides in the evaluation of pain in recent years, prompted a study in which Holt's work on the specificity of discography was replicated and extended. For the present study, seven patients who had low-back pain and ten volunteers who had been carefully screened, with a questionnaire and a physical examination, to ensure that they had no history of problems with the back, had an injection at three levels, and all sessions were videotaped. After each injection, the participant was interviewed about the pattern and intensity of the pain, and then the discs were imaged with computed tomography. Five raters, who were blind to the condition of the participant, graded each disc as normal or abnormal on the basis of findings on magnetic resonance images that had been made before the injection and computed tomography (discography) were done. There was only one disagreement between the ratings that were made on the basis of the magnetic resonance images and those that were made on the basis of the discograms.

Each participant's pain-related response was evaluated independently by two raters who viewed the videotapes of the discography. Inter-rater reliability was 0.99, 0.93, and 0.88 for the evaluation of intensity of the pain, pain-related behavior, and similarity of the pain to pain that the subject had before the injection. In the asymptomatic individuals, the discogram was interpreted as abnormal for 17 percent (5) of the 30 discs and for 5 of the 10 subjects. However, when the discogram was defined as positive only when substantial pain had been associated with the injection, there were no positive interpretations for the asymptomatic individuals (a false-positive rate of 0 and a specificity of 100 percent). Thus, although the prevalence of abnormal discograms in the asymptomatic participants in the present study was generally consistent with Holt's results, the false-positive rate in this study (0 percent) was significantly lower than in Holt's study (26 percent).

Discography revealed abnormal findings in 13 of the 20 discs in the symptomatic individuals and in all 7 patients

at one or more levels. Six patients had positive discograms on the basis of the study criteria. It was concluded that, with current techniques and in conjunction with standardized methods for assessment of pain, lumbar discography is a highly reliable and specific diagnostic test. This study did not address the validity or sensitivity or provide guidelines for choosing patients for discography. Certainly, discography is not the best diagnostic test for all patients who have low-back pain. [JEE]

Practical Methods for Controlling Powered Upper-Extremity Prostheses. Williams TW, reprinted from *Assist Technol* 2:3-18, 1990.

Myoelectric pickups (electrodes and processors for detecting the signal that is recorded as an electromyogram) are the most important human-machine interface for controlling powered upper-extremity prostheses. This article presents a simple explanation of myoelectric signal acquisition and then discusses how these signals are used to control the small motors in electric hands, elbows, wrist rotators, and other similar equipment. The less-familiar switch-based and proportional position-sensing controls are also explained. A complete listing of the major suppliers and products available will aid in understanding a discussion of the criteria for using external power instead of, or along with, body power to control and activate prosthetic function. [JEE]

Prosthetic Use and Functional and Social Outcome Following Major Lower Limb Amputation. Pohjolainen T, Alaranta H, Karkkainen M, reprinted from *Prosthet Orthot Int* 14:75-79, 1990.

A total of 175 consecutive below and above-knee amputees sent to the prosthetic workshop in Helsinki for prosthetic fitting from 32 hospitals were reviewed to determine their functional ambulation and social adaptation. The average age of patients was 62.2 years at the time of the prosthetic fitting. The mortality was 11 percent (19) during the first postoperative year. One-year postoperative information was obtained for 141 of the surviving patients (90 percent) by personal contact. At the time of the review, 68 percent of the amputees (96 patients) who had been fitted with a prosthesis made extensive and regular use of it. Half of all the above-knee amputees and 79 percent of the below-knee used their prosthesis throughout the day or over seven hours a day. A total of 72 percent of the above-knee amputees (33/46) and 85 percent of the below-knee amputees (67/79) had useful ambulation, at least indoors. Of the 141 patients contacted, 124 (88 percent) lived in their own homes. The remaining 16 patients (11 percent) lived in apartment houses for the aged or old people's homes. A total of 48 amputees (34 percent) needed a regular home help. [JEE]

Replantation of the Distal Part of the Leg. Usui M, Kimura T, Yamazaki J, reprinted from *J Bone Joint Surg* 72-A:1370-1373, 1990.

We successfully replanted five amputated legs in five patients and followed the patients for two years or more (average, six years and three months). Although some patients found it impossible to squat and to run because of joint contractures, muscle weakness, or deformities of the foot, all patients could perform other activities without difficulty. None had important pain or any intolerance to cold, and all were satisfied with the results of the replantation. [JEE]

The Role of the Contralateral Limb in Below-Knee Amputee Gait. Hurley GRB, McKenney R, Robinson, *et al.*, reprinted from *Prosthet Orthot Int* 14:33-42, 1990.

Very little quantitative biomechanical research has been carried out evaluating issues relevant to prosthetic management. The literature available suggests that amputees may demonstrate an asymmetrical gait pattern. Furthermore, studies suggest that the forces occurring during amputee gait may be unequally distributed between the contralateral and prosthetic lower limbs. This study investigates the role of the contralateral limb in amputee gait by determining lower limb joint reaction forces and symmetry of motion in an amputee and nonamputee population. Seven adult below-knee amputees and four nonamputees participated in the study. Testing involved collection of kinematic coordinate data employing a WATSMART video system and ground reaction force data using a Kistler force plate. The degree of lower limb symmetry was determined using bilateral angle-angle diagrams and a chain encoding technique. Ankle, knee and hip joint reaction forces were estimated in order to evaluate the forces acting across the joints of the amputee's contralateral limb. The amputees demonstrated a lesser degree of lower limb symmetry than the nonamputees. This asymmetrical movement was attributed to the inherent variability of the actions of the prosthetic lower limb. The forces acting across the joints of the contralateral limb were not significantly higher than that of the nonamputee. This suggests that, providing the adult amputee has a good prosthetic fit, there will not be

increased forces across the joints of contralateral limb and consequently no predisposition for the long-term wearer to develop premature degenerative arthritis. [JEE]

Shock Absorbing Material on the Shoes of Long Leg Braces for Paraplegic Walking. Bierling-Sorensen F, Ryde H, Bojsen-Moller F, Lyquist E, reprinted from *Prosthet Orthot Int* 14:27-32, 1990.

A study was designed to evaluate if shock absorbing material (ethyl vinyl acetate [EVA]) on the shoes of long leg braces could decrease the accelerations and consequent shock forces transmitted through the leg and brace during paraplegic walking. Six male paraplegic (26-55 years old) took part, four using a "swing-to" and two a "swing-through" technique when walking. Recordings comprised accelerometry of leg and brace, force platform measurement, and still photography of the trajectories of the leg segments. Each experimental condition was tested three times with a coefficient of variation (CV) for the measurements ranging from 5-22 percent. Compared to hard heels, shoes equipped with 20mm EVA soles decreased the acceleration amplitude in the first 10 msec as well as at maximum for shoe-to-ground contact. With the accelerometer at the malleolus reduction of the amplitude averaged 22 percent and 12 percent respectively, and 35 percent and 21 percent respectively with the accelerometer on the caliper (p: 0.03-0.1). In a second trail the two "swing-through" walkers had new shoes made with a 10mm thick EVA heel built in. After 3 months of walking with these shoes tests were carried out with the accelerometer attached to the malleolus both when the new and the former shoes were put on the calipers. CV for these measurements were 15-24 percent. It was found that the new shoes decreased the amplitudes by up to 62 percent and 26 percent on average (all $p < 0.01$). The experimental subjects indicated that the EVA soles/heels gave a more comfortable and silent walk, e.g., the "bump" transmitted up through the body to the head diminished. In future, shock absorbing material should be built into the heels of shoes provided to long leg braces for paraplegic walking. [JEE]

Use of a Knee-Brace for Control of Tibial Translation and Rotation: A Comparison, in Cadavera, of Available Models. Wojtys EM, Loubert PV, Samson SY, Viviano DM, reprinted from *J Bone Joint Surg* 72-A:1323-1329, 1990.

We assessed the relative restraints that are provided by fourteen currently available functional knee-braces, using six limbs in cadavera. The tests were made at 30 and 60 degrees of flexion of the knee, and a mechanical loading system applied loads that caused anterior-posterior translation and internal-external rotation. The braces limited abnormal tibiofemoral displacements by 10 to 75 percent in translation; there was more variation in rotation. This study demonstrated that functional knee-braces provide a restraining influence that may be beneficial in the control of abnormal displacements of the knee, but that the degree of restraint varies considerably. [JEE]

SENSORY AIDS/REHABILITATION

Acoustical Effects of Endotracheal Intubation. Yonick TA, Reich AR, Minifie FD, Fink BR, *J Speech Hear Disord* 55:427-433, 1990.

Examination of 13 postoperative males with 8 acoustical measures produced variable results. Predictable trends were identified. Monitoring even minor intubation-related laryngeal traumas may be successful using three of the eight measures. [JDS]

Characterization of Tinnitus by Tinnitus Patients. Stouffer JL, Tyler RS, *J Speech Hear Disord* 55:439-453, 1990.

Analysis of 528 tinnitus patients' responses to a questionnaire showed: 1) almost three of four patients reported it occurred 26 or more days per month; 2) for a little less than one in five patients tinnitus occurred in the presence of normal hearing or mild impairment; 3) one in four indicated their tinnitus had increased over time; 4) severity of tinnitus was greatest in those for whom it was the primary complaint and in those with Meniere's syndrome; and, 5) patients differed with respect to whether noise or quiet exacerbated their tinnitus. [JDS]

The Effect of Rate Control on the Intelligibility and Naturalness of Dysarthric Speech. Yorkston KM, Hammen VL, Deukelman DR, Traynor CD, *J Speech Hear Disord* 55:550-560, 1990.

Four patients with severe ataxic dysarthria and four with severe hypokinetic dysarthria read passages at rates controlled by the experimenters. Recordings of their productions were compared to those of four normal speakers and their own speech at their accustomed pace. Sentence intelligibility improved when rates were controlled for the

dysarthric patients. Slowing did not affect the naturalness of speech as much for the dysarthric as for normal individuals. Discrepancies occurred in measures of sentence and phoneme intelligibility. [JDS]

The Effectiveness of Six Insulin Management Devices for Blind Diabetic Persons. Ponchillia SV, Richardson K, Turner-Barry MM, *J Visual Impairm Blindn* 84:364-370, 1990.

The devices tested were: Click-Count syringe, Dos-Aid, Inject-Aid, Meditec Insulgage, Orange Medical Instruments' Betafect, and Squibb-Novo NovoPen. Using a single-subject design, three people evaluated each device. Two subjects preferred NovoPen and one the Insulgage. Each device has its advantages and disadvantages. Suggestions for further research are offered. [JDS]

Estimating Respiratory Volumes from Rib Cage and Abdominal Displacements during Ventilatory and Speech Activities. Reich AR, McHenry MA, *J Speech Hear Res* 33:467-475, 1990.

After practice, 10 normal males, with mesomorphic body types, performed two tests of vital capacity, read the Rainbow passage, and submitted to measurements of tidal ventilation. The results indicate the problems inherent in relating chest-wall kinematics to exchanged volumes during ventilation and speech. The authors believe that "appropriateness of using respiratory kinematic devices as calibrated volume instruments obviously depends on the nature of the research or clinical questions that one is asking...." [JDS]

Evaluation of an In-Situ Output Probe-Microphone Method for Hearing Aid Fitting Verification. Cox RM, Alexander GC, *Ear Hear* 11(1):31-39, 1990.

The article describes the use of a probe microphone to verify hearing-aid fitting in patients with sensorineural impairments. Each of 24 patients had three aids prescribed, using: 1) MSUv3; 2) MSUv3 plus 4 dB per octave; and, 3) MSUv3 minus 4 dB per octave. For over-the-ear hearing aids, prescriptions were matched with a 5 dB RMS error or less. The probe-microphone procedure yielded accurate fittings. However, it required precise control of the simulated speech stimuli and other inputs. [JDS]

A Longitudinal Study of Depression Among Stroke Patients Participating in a Rehabilitation Program. Bacher Y, Korner-Bitensky N, Mayo N, Becker R, Coopersmith H, *Can J Rehabil* 4:27-37, 1990.

As common as depression is following stroke, little is empirically known about its course over a substantial time period. In the year they were followed post-stroke, 48 patients exhibited the high amount of depression anticipated. The prevalence of depression increased for males and remained the same for females. Their functioning improved in all cases, but to a greater extent among those without accompanying depression. The authors urge aggressive treatment of the emotional concomitants of stroke. [JDS]

The Modified Tongue Anchor Technique as a Screening Test for Velopharyngeal Inadequacy: A Reassessment. Dalston RM, Warren DW, Dalston ET, *J Speech Hear Disord* 55:510-515, 1990.

Assessment of 311 consecutive admissions out of 900 (exclusions based on not meeting criteria, such as age) showed that the screening test had a specificity of 0.85 and a sensitivity of 0.78. Despite the respectable validities of the test, the authors caution that "it certainly would be inappropriate to use the tongue-anchor task... as an isolated tool for assessing a patient's ability to close off the nose from the mouth during speech." [JDS]

Performance of Cochlear Implant Patients as a Function of Time. Spivak LG, Waltzman SB, *J Speech Hear Res* 33:511-519, 1990.

On the average, 15 deafened adults' speech perception improved over a three-year period following implantation with the Nucleus 22. The greatest comparative improvement occurred between the pre-implant and three months post-implant periods, but some patients showed improvements into the third year as measured by MAC scores. Improvement related to the processing scheme used by the patient: those that included coding F1 showed significantly greater improvement than F0F2 only. A limiting factor in generalizing these findings is the MAC's test-retest characteristics, which are presently unknown. Thus, whether results of this research show people learn to perform the MAC more successfully over time or make better use of their implant's information remains open to question. [JDS]

Preferred Insertion Gain of Hearing Aids in Listening and Reading-Aloud Situations. Kuk FK, *J Speech Hear Res* 33:520-529, 1990.

Comparing a group of nine persons with high-frequency hearing impairments to eight with high- and low-frequency impairments, authors found both groups preferred more gain when listening to other persons' voices than their own. Authors believe multimemory hearing aids may be necessary for some patients. [JDS]

Prevalence and Causes of Blindness and Visual Impairment in Canada. Naeyaert KM, Grace G, *J Visual Impairm Blindn* 84:361-363, 1990.

Persons identified in the 1986 Canadian Census of the Population as having a limitation of activity or health condition were followed up by the Health and Activity Limitation Survey (HALS). HALS also sampled a large portion of those who responded no to the census screening questions. HALS estimated 445,875 Canadian adults have impaired vision; i.e., trouble reading ordinary newsprint with glasses, if normally worn, and/or trouble seeing clearly the face of someone 12 feet across a room, also with glasses, if normally worn. Among the findings: 1) as age increased, so did visual impairment; and, 2) HALS' estimated rate for legal blindness was far lower than the Canadian National Institute for the Blind's; self-assessed degree of impairment was related to reported cause. [JDS]

A Quick Look at the NOMAD: An Audio-Tactile Graphics Processor. Uslan M, Schreier E, Meyers A, *J Visual Impairm Blindn* 84:383-384, 1990.

Describes NOMAD, an adaptation of a graphic tablet used with an IBM PC. It enables the blind person to read a figure, such as a map, by supplementing its tactile representation with speech. It has "great potential as a teaching aid for a variety of tactile and visual graphics. The technology itself is basically well designed." Details with respect to sources, prices, etc., are also provided. [JDS]

Reducing Late-Life Dependence Resulting from Declining Visual Activity. Vaughan CE, Hobson S, *J Visual Impairm Blindn* 84:364-370, 1990.

Publicity attracted 3,393 elderly persons from 39 predominantly rural counties of Missouri to visual screening. Low-vision centers were located at sites near large concentrations of older persons. The authors were surprised by the "high level of use of and the important contribution that these vision centers made to the residents and staffs of the participating nursing homes." The study concluded that visual impairments were extensive among elderly people, that low-vision aids are useful, and that rehabilitation services are scarce in the areas studied. [JDS]

The Relationship Between Psychopathology and Speech and Language Disorders in Neurologic Patients. Sapir S, Aronson AE, *J Speech Hear Disord* 55:503-509, 1990.

The authors review the literature relating psychopathology to speech and language disorders. The importance of anxiety, conversion reactions, and depression in conditions like aphasia, apraxia, dementia, and voice disorders should be considered by speech pathologists when treating patients with communication problems. The effects of behavioral modification, drugs, and other treatments should be assessed with affective factors clearly in the pathologist's view. [JDS]

Selected Acoustic Characteristics of Voices Before Intubation and After Extubation. Horii Y, Fuller BF, *J Speech Hear Res* 33:505-510, 1990.

Various measures of voice characteristics of eight males and eight females who had undergone endotracheal intubation led authors to conclude that: 1) the procedure affects the acoustic characteristics of speech even after relatively short durations; and, 2) measures of waveforms are sufficiently sensitive to detect the vocal alterations. [JDS]

Semantic Memory Deterioration in Alzheimer's Subjects: Evidence from Word Association, Definition, and Associate Ranking Tasks. Abeysinghe S, Bayles KA, Trosset MW, *J Speech Hear Res* 33:574-582, 1990.

Three cognitive tasks were administered to 23 Alzheimer's patients. They gave more repetitious, multiword, and unrelated responses to word associations than normal controls and lower ratios of paradigmatic to syntagmatic responses. Other associative tasks were similarly depressed. However, patients defined many words for which they did not give meaningful word associations. [JDS]

What Is Stuttering? Perkins WH, *J Speech Hear Disord* 55:370-382, 1990.

Author proposes a definition of stuttering that is independent of the fact that listeners cannot acceptably judge unit-by-unit occurrences of it. "My hypothesis is that what is frustrating, and can become frightening, about stuttering is that when it happens speech is either about to be or becomes temporarily disrupted for reasons the stutterer is unable to prevent. Moreover, the disruption is specific to the speech attempt." Note that four subsequent articles in this issue respond to this paper, along with a rebuttal by Perkins. [JDS]

BOOK REVIEWS

by
Jerome D. Schein, Ph.D.
Professor Emeritus of Senory Rehabilitation, New York University, and David Peikoff Chair in Deafness Studies, University of Alberta, Edmonton, Canada

AIDS and Deafness Resource Directory, 2nd edition, by National AIDS Information Clearinghouse. Atlanta, GA: Center for Disease Control, 1989. 105 pp.

The directory provides information about national, state, and local facilities offering AIDS-related services to deaf and hard of hearing persons. To be included, an organization must have had one or more of the following: TDD access, sign-language interpreters, and/or educational materials geared to the reading levels of the average deaf person. By its publication, this document comments on rehabilitation in this era.

Demographic and Large-Scale Research with Hearing-Impaired Populations: An International Perspective, edited by Amatzia Weisel. Washington, DC: Gallaudet Research Institute, Gallaudet University; Rochester, NY: National Technical Institute for the Deaf, University of Rochester, 1990. 162 pp. Illustrated.

Few rehabilitationists need to be convinced about the value of statistics on the various conditions with which they deal. This monograph addresses the methodological aspects of gathering data on persons with impaired hearing. It also contains a variety of current information about hearing-impaired populations in several countries: Canada, Great Britain, Israel, West Germany, and the United States. One chapter presents data from a European survey of interpreter services for deaf people, further enhancing the monograph's claim to an "international perspective." The 14 chapters were originally oral presentations, but the editor wisely granted each participant the privilege of redoing them for publication, thus avoiding the annoying impression usually left by speeches gathered into hard covers. Since one of the chapters is by myself, I will eschew evaluative comments. The description alone should suffice to encourage readers to obtain this document and, then, to form an opinion of their own as to its value.

How to Survive Hearing Loss, by Charlotte Himber. Washington, DC: Gallaudet University Press, 1989. 241 pp. Foreword by Richard Dysart.

Rehabilitationists seeking readings for newly hard of hearing and deafened patients will be pleased by Himber's well-written account of her own impairment. As a former editor, she organizes material clearly and presents it in a highly readable style that will appeal to lay readers. By using her own experiences, the author does much to overcome the initial shock and subsequent stigma that typically accompanies the onset of impaired hearing. At the same time, professionals need not be concerned that she has departed so far from established practice and the current state of knowledge that they will have to spend hours correcting misapprehensions—a failing of some other lay-directed books. Her book carries the headnote: "This book is not intended as a substitute for the medical advice of a physician." Would that more self-help materials were as conscientious in providing similar cautions up front.

Older Americans and Tinnitus: A Demographic Study and Chartbook, GRI Monograph Series A, No. 2, by Scott Campbell Brown. Washington, DC: Gallaudet Research Institute, Gallaudet University, 1990, 97 pp. Illustrated.

Audiologists and otologists eager for statistics about this highly prevalent symptom will find great satisfaction in this carefully researched, excellently presented monograph, supplemented by a 57-item bibliography. Whether one accepts a prevalence rate among U.S. adults of 31.4 percent or 2.6 percent, one must be impressed by the widespread nature of this affliction. Brown has put together an impressive array of data and advances six hypotheses to account for the discrepant statistics. His major sources of information are the Health Interview Surveys of 1982 through 1987, the 1984 Supplement on Aging, and the

1971-1975 Health and Nutrition Examination Survey—all conducted by the National Center for Health Statistics. In addition to the prevalence rates, the monograph presents an impressive array of related data: demographic, socio-economic, auditory, and health characteristics. Brown does not confuse the numerous associations between tinnitus and these variables as necessary or sufficient conditions; he wisely concludes that "the prevalence of tinnitus does not have to increase in the future, if the problems of hearing impairment and poor health among older Americans are prevented."

PUBLICATIONS OF INTEREST

Compiled by Beryl M. Benjers, Ph.D.
Departments Editor

This list of references offers *Journal* readers significant information on the availability of recent rehabilitation literature in various scientific, engineering, and clinical fields. The *Journal* provides this service in an effort to fill the need for a comprehensive and interdisciplinary indexing source for rehabilitation literature.

All entries are numbered so that multidisciplinary publications may be cross-referenced. They are indicated as *See also* at the end of categories where applicable. A listing of the periodicals reported on follows the references. In addition to the periodicals covered regularly, other publications will be included when determined to be of special interest to the rehabilitation community. To obtain reprints of a particular article or report, direct your request to the appropriate contact source listed in each citation.

Page **List of Categories**

BIOENGINEERING and BIOMECHANICS

1. Application of Computer-Assisted Engineering (CAE) Software to Orthopedic Surgical Planning. Sutherland CJ, *et al.*, *SOMA Eng Hum Body* 3(4):49-56, 1990.
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Contact: A.O. Andrisano, Faculty of Engineering, University of Modena, Italy

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302. A New Racing Wheelchair Front Wheel Mounting Mechanism. Cooper RA, *Palaestra* 7(1):30-32, 1990.
Contact: Rory A. Cooper, PhD, Biomedical Engineering Program, Dept. of Electrical Engineering, California State University, Sacramento, CA 95819

303. Seat Height in Hand Rim Wheelchair Propulsion: A Follow-Up Study. van der Woude L, *et al.*, *J Rehabil Sci* 3(3):79-83, 1990.
Contact: Luc van der Woude, Faculty of Human Movement Sciences, Free University, Amsterdam, The Netherlands

See also 178

WOUNDS and ULCERS

304. The Effects of Age and Peripheral Vascular Disease on the Circulatory and Mechanical Response of Skin

to Loading. Czerniecki JM, *et al.*, *Am J Phys Med Rehabil* 69(6):302-306, 1990.
Contact: Joseph M. Czerniecki, MD, VA Medical Center, RMS (117), 1660 S. Columbian Way, Seattle, WA 98108

305. The Management of Burns (Review article). Deitch EA, *N Engl J Med* 323(18):1249-1253, 1990.
Contact: Edwin A. Deitch, MD, Division of Burns and Trauma, Louisiana State University Medical Center, Shreveport, LA 71130

306. Pressure Ulcers: Prevention is the Key. Randall C, *Paraplegia News* 44(9):38-39, 1990.
Contact: Carrie Randall, RN, Spinal Cord Injury and Psychiatry, VA Medical Center, 10701 East Blvd., Cleveland, OH 44106

307. The Role of Alternating Air and Silicore® Overlays in Preventing Decubitus Ulcers. Conine TA, Daechsel D, Lau MS, *Int J Rehabil Res* 13(1):57-65, 1990.
Contact: Tali A. Conine, Dept. of Rehabilitation Medicine, University of British Columbia, Vancouver, British Columbia V6T 1W5, Canada

See also 35, 265

Periodicals reviewed for PUBLICATIONS OF INTEREST

Accent on Living
Acta Orthopaedica Scandinavica
Advances in Orthopaedic Surgery
American Annals of the Deaf
American Journal of Occupational Therapy
American Journal of Physical Medicine and Rehabilitation
American Journal of Sports Medicine
American Rehabilitation
Annals of Biomedical Engineering
AOPA Almanac (American Orthotic and Prosthetic Association)
Applied Optics
Archives of Physical Medicine and Rehabilitation
ASHA (American Speech and Hearing Association)
Assistive Technology
Biomaterials, Artificial Cells and Artificial Organs
Biomedical Instrumentation & Technology
British Journal of Occupational Therapy
Caliper (Canadian Paraplegic Association)
Canadian Journal of Occupational Therapy
Canadian Journal of Rehabilitation
Clinical Biomechanics
Clinical Kinesiology
Clinical Orthopaedics and Related Research
Clinical Physics and Physiological Measurement
Clinical Rehabilitation
Communication Outlook
CRC Critical Reviews in Biomedical Engineering
DAV Magazine (Disabled American Veterans)
Discover
Electromyography and Clinical Neurophysiology
Electronic Design
Electronic Engineering
Electronics
Ergonomics
Harvard Health letter
Headlines: The Brain Injury Magazine
Hearing Journal
Hearing Research
Human Factors: The Journal of the Human Factors Society
IEEE Engineering in Medicine and Biology Magazine
IEEE Transactions in Biomedical Engineering
IEEE Transactions in Systems, Man and Cybernetics
International Disability Studies
International Journal of Rehabilitation Research
International Journal of Technology & Aging
JAMA
Journal of Acoustical Society of America
Journal of American Optometric Association
Journal of Association of Persons with Severe Handicaps
Journal of Biomechanical Engineering
Journal of Biomechanics
Journal of Biomedical Engineering
Journal of Biomedical Materials Research
Journal of Bone and Joint Surgery—American Ed.
Journal of Bone and Joint Surgery—British Ed.
Journal of Clinical Engineering
Journal of Head Trauma and Rehabilitation
Journal of Medical Engineering and Technology
Journal of Neurologic Rehabilitation
Journal of Optical Society of America A
Journal of Orthopaedic and Sports Physical Therapy
Journal of Orthopaedic Research
Journal of Prosthetics and Orthotics
Journal of Rehabilitation
Journal of Rehabilitation Sciences
Journal of Speech and Hearing Research
Journal of Trauma
Journal of Vision Rehabilitation
Journal of Visual Impairment and Blindness

Journal of Rehabilitation Research and Development Vol. 28 No. 2 Spring 1991

Laser Focus World
Mayo Clinic Proceedings
Medical and Biological Engineering and Computing
Medical Device and Diagnostic Industry
Medical Electronics
Medical Physics
Medical Progress Through Technology
Medical Psychotherapy Yearbook
Medicine & Science in Sports and Exercise
Military Medicine
New England Journal of Medicine
The Occupational Therapy Journal of Research
Optometry and Vision Science
Orthopaedic Review
Orthopedic Clinics of North America
Orthopedics
Palaestra
Paraplegia
Paraplegia News
Physical and Occupational Therapy in Geriatrics
Physical Medicine and Rehabilitation
Physical Therapy
Physics Today
Physiotherapy
Proceedings of the Institution of Mechanical Engineers—Part H: Journal of Engineering in Medicine
Prosthetics and Orthotics International
Rehab Management
Rehabilitation Digest
Robotics World
Scandinavian Journal of Rehabilitation Medicine
Science
Science News
Scientific American
SOMA: Engineering for the Human Body
Speech Technology
Spine
Sports 'N Spokes
Technical Aid to the Disabled Journal
Techniques in Orthopaedics
Topics in Geriatric Rehabilitation
VA Practitioner
Vanguard
Volta Review
Worklife

CALENDAR OF EVENTS

Compiled by Beryl M. Benjers, Ph.D.
Departments Editor

NOTE: An asterisk at the end of a citation indicates a new entry to the calendar.

1991

May 1-5, 1991
American Back Society Spring Symposium on Back Pain, Toronto, Ontario, Canada
Contact: Aubrey A. Swartz, MD, American Back Society, 2647 East 14th St., Suite 401, Oakland, CA 94601; (415) 536-9929, Fax (415) 536-1812

May 2-4, 1991
Second Biennial World Congress of the International Association for the Study of Traumatic Brain Injury: Technology in Brain Injury Management, Oslo, Norway
Contact: Henry H. Stonnington, MD, Sheltering Arms Rehab Hospital, 1311 Palmyra Ave., Richmond, VA 23227; Fax (804) 379-0828, or Arnstein Finset, PhD, Sunnaas Rehabilitation Hospital, 1450 Nesoddtangen, Norway*

May 2-5, 1991
American Orthotic and Prosthetic Association (AOPA), Region II & III Meeting, Williamsburg, VA
Contact: John Schulte, CPO, (301) 224-2874*

May 3-4, 1991
Reflex Sympathetic Dystrophy: Current Strategies in Diagnosis and Treatment, Philadelphia, PA
Contact: Office of Continuing Medical Education, Jefferson Medical College, 1025 Walnut St., Philadelphia, PA 19107-5083; (215) 955-6992

May 4-7, 1991
6th Annual National Symposium on Information Technology (NSIT), Myrtle Beach, SC
Contact: NSIT, The University of South Carolina, Columbia, SC 29208; (803) 777-4435

May 5-9, 1991
4th International Pre-Prosthetic Surgery Conference, Adelaide, Australia
Contact: Multinational Meetings Information Service BV, J.W. Brouwersplein 27, PO Box 5090, 1007 Amsterdam, The Netherlands*

May 5-10, 1991
Neural Networks: From Foundations to Applications (course), Tyngsboro, MA
Contact: Neural Networks, Wang Institute of Boston University, 72 Tyng Rd., Tyngsboro, MA 01879*

May 8-12, 1991
Orthopaedic and Rehabilitation Technology Trade Fair and Congress, Berlin, Germany
Contact: AMK Berlin Ausstellungs-Messe-Kongress-GmbH, Messedamm 22, D-1000 Berlin 19, Germany*

May 9-12, 1991
American Board of Physical Medicine and Rehabilitation, Rochester, MN
Contact: American Board of Physical Medicine and Rehabilitation, Suite 674, Northwest Center, 21 First St. SW, Rochester, MN 55902

May 9-12, 1991
American Geriatrics Society, Chicago, IL
Contact: American Geriatrics Society, Suite 400, 770 Lexington, New York, NY 10021

May 9-12, 1991
American Orthotic and Prosthetic Association (AOPA), Region IX Annual Meeting, High Sierra, South Lake Tahoe, CA
Contact: Lynn Crotto, (415) 621-4244*

May 10-12, 1991
Neural Networks for Vision and Image Processing, Tyngsboro, MA

Contact: Neural Networks, Wang Institute of Boston University, 72 Tyng Rd., Tyngsboro, MA 01879*

May 10-14, 1991
53rd Annual Meeting of the American Fracture Society, Washington, DC
Contact: John L. Wright, MD, PO Box 668, Bloomington, IL 61702-0668; (309) 663-6272*

May 11-15, 1991
26th Annual Meeting of the Association for the Advancement of Medical Instrumentation (AAMI), Washington, DC
Contact: AAMI, 3330 Washington Blvd., Suite 400, Arlington, VA 22201-4598*

May 13-16, 1991
Meeting of the International Society for the Study of the Lumbar Spine, Heidelberg, Germany
Contact: Prof. Alf Nachemson, Dept. of Orthopaedics, Sahlgren Hospital, S-413 45 Goteborg, Sweden*

May 14-17, 1991
VIth International Symposium on Audiological Medicine, Pecs, Hungary
Contact: Judith Burja, Cooptourist, H-1012 Budapest, Logodi u. 30, Hungary*

May 15-17, 1991
1991 Annual Scientific Meeting of the International Medical Society of Paraplegia (IMSOP), Stoke Mandeville Hospital, England
Contact: The Honorary Secretary, IMSOP, National Spinal Injuries Centre, Stoke Mandeville Hospital, Aylesbury, Bucks HP21 8AL, UK; Tel 0296 84111

May 16-18, 1991
American Orthotic and Prosthetic Association (AOPA), Region V Annual Meeting, Columbus, OH
Contact: Don Peters, (419) 522-0055

May 17-19, 1991
Prescription Footwear Association (PFA), "Advanced Pedorthic Course," Akron, OH
Contact: PFA, 9861 Broken Land Pkwy., Suite 255, Columbia, MD 21046; (800) 673-8447*

May 18-21, 1991
ISPO Course on Lower Limb Amputations and Related Prosthetics, Port el Kantaoui, Tunisia
Contact: ISPO, Borgervaenget 5, 2100 Copenhagen o, Denmark*

May 19-22, 1991
National Council on Aging (NCA), Miami, FL
Contact: NCA, 600 Maryland Ave. SW, West Wing 100, Washington, DC 20024

May 20-22, 1991
Seventh Oswestry Spine Symposium, Oswestry, Shropshire, UK
Contact: Erica Wilkinson, Symposium Secretary, The Robert Jones & Agnes Hunt Orthopaedic Hospital, Oswestry, Shropshire SY10 7AG, UK; Tel 0691 655311, Ext 3392*

May 22-24, 1991
President's Committee on Employment of People with Disabilities (PCEPD) Annual Meeting, "Work: The Key to Opportunity," Dallas, TX
Contact: Edmond Leonard, PCEPD, 1111 20th St. NW, Suite 636, Washington, DC 20036-3470; (202) 653-5044*

May 23-24, 1991
Netherlands Society of Rehabilitation and Physical Medicine Postgraduate Course: Rehabilitation of Physically Impaired and Disabled Children, Nijmegen, The Netherlands
Contact: Dr. H.G. Berghauser Pont, Bio Rehabilitation Centre for Children, PO Box 9215, 6800 HT Arnhem, The Netherlands; Tel 08308-38911*

May 23-25, 1991
IVth International Symposium of Advances in Orthopaedics and Trauma: Reconstructions of Large Bone Defects, Madrid, Spain
Contact: DNA, Angela Duran, Secretaria Traumatologia, Planta 5a Sur., Hospital Universitario San Carlos, Cuidad Universitaria, 28040 Madrid, Spain; Tel 34 1 2441500 or 2441705, Ext 253; Fax 549266*

May 26-29, 1991
5th Canadian Congress of Rehabilitation: Science, Dignity, Opportunity, Charlottetown, Prince Edward Island
Contact: Congress Secretariat, CRCD, 45 Sheppard Ave. E, Suite 801, Toronto, Ontario M2N 5W9, Canada; (416) 250-7490, Fax (416) 229-1371

May 26-31, 1991
Deaf Worlds: Together Towards 2000, Eastbourne, Sussex, UK

Contact: RNID Conference Organizer, Conference Associates and Services Ltd., 55 New Cavendish St., London W1M 7RE, England; Fax 071-935-7559 or Royal National Institute for the Deaf, 105 Gower St., London WC1E 6AH, England*

May 27-31, 1991
IVth International Symposium on Quantitative Luminescence Spectrometry in Biomedical Sciences, Ghent, Belgium
Contact: State University of Ghent, Pharmaceutical Institute, Harelbekestraat 72, B-9000 Ghent, Belgium*

June 1-5, 1991
American Occupational Therapy Association (AOTA), 71st Annual National Conference and Exposition, Cincinnati, OH
Contact: AOTA, 1383 Piccard Dr., Rockville, MD 20850-4375; (301) 948-9626*

June 2-5, 1991
The Fourth International Conference on Industrial and Engineering Applications of Artificial Intelligence and Expert Systems, IEA/AIE '91, Waiohai, Hawaii
Contact: Sandra Shankle, IEA/AIE '91, The University of Tennessee Space Institute, Tullahoma, TN 37388-8897; (615) 455-0631, Ext 278

June 3-6, 1991
American Orthopaedic Association (AOA) Annual Meeting, Palm Beach, USA
Contact: AOA, 222 S. Prospect Ave., Park Ridge, IL 60068

June 3-7, 1991
Netherlands Society of Rehabilitation and Physical Medicine Postgraduate Course: Examination of the Postural and Motor System, Introduction to Scientific Research, The Netherlands
Contact: Prof. W.H. Eisma, Rehabilitation Dept., University Hospital, PO Box 30001, 9700 RB Groningen, The Netherlands; Tel 050-619111*

June 5-8, 1991
American Medical Informatics Association (AMIA), Second Annual Education and Research Conference: The Unified Medical Language System Informatics and Health Outcomes, San Francisco, CA
Contact: AMIA, 11140 Rockville Pike, Box 324, Rockville, MD 20852

June 6-8, 1991
5th European Congress of the Society for Surgery of the Shoulder and the Elbow, Wurtzburg, Germany
Contact: Prof. Dr. med. J. Eulert, Orthopadische Universitatsklinik, Konig-Ludwig-Haus, Brettreich-str. 11, 8700 Wurtzburg, Germany; Tel 0931 80 32 12; Fax 0931 80 32 25*

June 6-9, 1991
Back Pain-Current Concepts and Recent Advances: Fourth International Meeting, Calgary, Canada
Contact: The Secretariat, 30 Deane Way, Ruislip, Middlesex HA4 8SX, UK*

June 6-9, 1991
International Back Pain Society, Sydney, Australia
Contact: The Secretariat, Congress Team International (UK) Ltd., 30 Deane Way, Ruislip, Middlesex, UK*

June 10-13, 1991
Call for Papers: ISPO Scientific Seminar on Clinical Biomechanics of Foot and Shoe, Jonkoping, Sweden
Contact: Secretariat of ISPO Scientific Seminar, Attn: Ulla-Britt Johansson, Dept. of Biomechanics and Orthopedic Technology, University College of Health and Care, PO Box 1038, S-551 11 Jonkoping, Sweden; Tel 46 36 10 47 53, Fax 46 36 10 47 62*

June 10-14, 1991
SYSTED 1991 Systems Science in Health and Social Services for the Elderly and the Disabled (Fourth International Conference), Barcelona, Spain
Contact: Inter-Congress, Gran Via de los Corto, Catalanes, 646, 08007, Barcelona, Spain*

June 12-16, 1991
American Orthotic and Prosthetic Association (AOPA), Quad-Regional Meeting, New Orleans, LA
Contact: Henry Lambert, (504) 344-1533*

June 16-19, 1991
1991 Biomechanics Symposium, ASME Applied Mechanics and Biomechanics Conference, Columbus, OH
Contact: Robert L. Spilker, ScD, Dept. of Mechanical Engineering, Aeronautical Engineering, and Mechanics, Rensselaer Polytechnic Institute, Troy, NY 12180-3590; (518) 276-6981

June 17-20, 1991
Canadian Organization of Medical Physicists Annual Meeting with Canadian College of Physicists in Medicine and Canadian Radiation Protection Association, Winnipeg, Manitoba, Canada
Contact: Dr. Walter Huda, Medical Physics, 100 Olivia St., Winnipeg, Manitoba R3E 0V9, Canada; (204) 787-4191 or Fax (204) 783-6875

June 19-20, 1991
ICORR '91, 1991 International Conference on Rehabilitation Robotics, Atlanta, GA
Contact: Michael Burrow, (404) 894-7034

June 20-22, 1991
'91 ICAR, Fifth International Conference on Advanced Robotics, Pisa, Italy
Contact: '91 ICAR Secretariat, Consorzio Pisa Ricerche, Via Risorgimento, 9, I-56126 Pisa, Italy; Tel +39 50 500995/502015, Fax +39 50 501016

June 20-23, 1991
American Orthotic and Prosthetic Association (AOPA), Region VI Annual Meeting, Indianapolis, IN
Contact: Angela Brenner, (414) 445-8840*

June 21-23, 1991
International Hearing Aid Conference: Signal Processing, Fitting, & Efficacy, Iowa City, IA
Contact: Dept. of Speech Pathology and Audiology, University of Iowa, Iowa City, IA; (319) 335-8726, Fax (319) 335-8851*

June 21-26, 1991
RESNA Rehabilitation Technology, 14th Annual Conference, Kansas City, MO
Contact: RESNA, Association for the Advancement of Rehabilitation Technology, Suite 700, 1101 Connecticut Ave. NW, Washington, DC 20036; (202) 857-1199

June 23-27, 1991
American Physical Therapy Association, Annual Conference, Boston, MA
Contact: Bonnie Polvinale, APTA, 1111 N. Fairfax St., Alexandria, VA 22314; (703) 684-2782*

June 24-28, 1991
IAPSRS Conference-Psychosocial Rehabilitation: An *Emphasis on Outcome*, *Baltimore*, MD
Contact: Jean Taylor, Conference Program Coordinator, MAPS, 20 Winters Lane, 2nd Floor, Catonsville, MD 21228*

June 26-29, 1991
Cervical Spine Research Society: 3rd Common Meeting, Athens, Greece
Contact: Scientific Committee, 3rd Common Meeting of the CSRS, c/o Prof. Dem. St. Korres, 10 Heyden St., 10434 Athens, Greece*

June 26-30, 1991
Fifty-Fifth Biennial Meeting of the Convention of American Instructors of the Deaf and the Conference of Educational Administrators Serving the Deaf, New Orleans, LA
Contact: Diane H. Bordelon or Connie Tullos, Louisiana School for the Deaf, PO Box 3074, Baton Rouge, LA 70821; (504) 769-8160, Ext 211 (Voice or TDD)*

June 27-28, 1991
Society for Research in Rehabilitation (SRR), Summer Meeting on Provision of Rehabilitation Services: Models for 1990s, Bristol, UK
Contact: SRR Organising Group, Dept. of Neurology and Stroke Rehabilitation Unit, Frenchay Hospital, Bristol BS16 1LE, UK*

July 3-5, 1991
Advances in Hydrotherapy II: Movement and Immersion in Water, Nijmegen, The Netherlands
Contact: Congress Office, c/o Mrs. J. Koot, University of Nijmegen, PO Box 9111, 6500 HN Nijmegen, The Netherlands; Fax 318 056 7956

July 7-12, 1991
World Congress on Medical Physics and Biomedical Engineering, Kyoto, Japan
Contact: Japan Convention Services Inc., Osaka Branch, Sumitomo Seimei Midosuji Bldg., 4-14-3, Nishitemma, Kita-ku, Osaka 530, Japan; Tel 06-311-3131, Fax 06-311-2130

July 8-12, 1991
Thirteenth Annual Diagnostic Imaging Seminar: A Practical Analysis of Various Imaging Procedures, Martha's Vineyard, MA
Contact: Nancy Fedullo, Dept. of Radiology, Hospital of the University of Pennsylvania, 3400 Spruce St., Philadelphia, PA 19104; (215) 662-6982*

July 10-12, 1991
ConnSENSE '91: Eighth Annual Conference on Technology for Individuals with Disabilities, Storrs, CT
Contact: Chauncy N. Rucker, University of Connecticut Special Education Technology Lab, 249 Glenbrook Rd., U-64, Storrs, CT 06269-2064; (203) 486-0165*

July 21-25, 1991
American Association of Physicists in Medicine (AAPM), 33rd Annual Meeting, San Francisco, CA
Contact: AAPM, 335 East 45th St., New York, NY 10017; (212) 661-9404

July 22-25, 1991
Engineering and Physical Sciences in Medicine, Sydney, Australia
Contact: Christine Bourke, International Professional and Continuing Education Institute, University of New South Wales, Kensington, NSW 2033, Sydney, Australia, Tel 02 6973178*

July 22-26, 1991
Clinical Magnetic Resonance Imaging: Principles and Applications (Course), Southhampton Princess, Bermuda
Contact: Nancy Fedullo, Dept. of Radiology, Hospital of the University of Pennsylvania, 3400 Spruce St., Philadelphia, PA 19104; (215) 662-6982*

July 23-26, 1991
Pre-Congress Course for Physiotherapists, Oswestry, Shropshire, UK
Contact: Erica Wilkinson, Symposium Secretary, The Robert Jones & Agnes Hunt Orthopaedic Hospital, Oswestry, Shropshire SY10 7AG, UK; Tel 0691 655311 Ext 3392*

July 28-July 31, 1991
International Symposium on 3D Analysis of Human Movement, Montreal, Canada
Contact: Secretariat, International Symposium on 3D Analysis of Human Movement, Laboratoire d'Etude du Mouvement, Centre de Recherche Pediatrique, Hopital Saint-Justine, 3175 Cote Ste-Catherine, Montreal, Quebec H3T 1C5, Canada*

July 28-August 1, 1991
33rd Annual Meeting: American Association of Physicists in Medicine (AAPM), New York, NY
Contact: AAPM, 335 East 45th St., New York, NY 10017*

July 28-August 2, 1991
11th Congress of the World Confederation for Physical Therapy, London, England
Contact: Secretariat, WCPT, Conference Associates, 27A Medway St., London SW1P 2BD, England*

August 5-7, 1991
4th European Congress on Research in Rehabilitation, Ljubljana, Yugoslavia
Contact: University Rehabilitation Institute, Linhartova 51, 61000, Ljubljana, Yugoslavia; Tel 3861 329 366, Fax 3861 329 366 or 324 392, Telex 32260

August 7-11, 1991
Southern Orthopaedic Association (SOA) Meeting, Colorado Springs, CO
Contact: SOA, 222 S. Prospect Ave., Park Ridge, IL 60068*

August 18-23, 1991
Fifth Congress of the International Psychogeriatric Association (IPA), Jerusalem, Israel
Contact: IPA Fifth Congress, PO Box 50006, Tel Aviv 61500, Israel

August 19-22, 1991
VIII World Congress of the International Society for Artificial Organs, Montreal, Quebec, Canada
Contact: Prof. T.M.S. Cheng, Artificial Cells and Organs Research Centre, Faculty of Medicine, McGill University, 3655 Drummond St., Montreal, Quebec H3G 1Y6, Canada; (514) 398-3512; Fax (514) 398-4983

September 1-6, 1991
6th World Congress in Ultrasound, Copenhagen, Denmark
Contact: 6th World Congress in Ultrasound, Spadille Congress Service, Sommervej 3, DK 3100 Hornbaek, Denmark*

September 2-4, 1991
Accessibility Legislation, Expert Seminar and Workshop, Budapest, Hungary
Contact: CIB W84 Secretariat, Dept. of Building Function Analysis, The Royal Institute of Technology, 100 44 Stockholm, Sweden; Fax 46 8 790839

September 2-6, 1991
6th Meeting World Federation for Ultrasound in Medicine and Biology, Copenhagen, Denmark
Contact: Soren Hanke, Ultralydlaboratoriet, Kobenhavns Amts Sygehus, Gentofte, DK-2900 Hellerup, Denmark

September 4-6, 1991
European Conference: New Initiatives in Traumatic Brain Injury Rehabilitation for Adults, Leeds, UK
Contact: Prof. M.A. Chamberlain, 36 Clarendon Rd., Leeds LS2 9NZ, UK; Tel (0532) 334936/334935, Fax (0532) 336017*

September 6-7, 1991
American Academy of Orthotists and Prosthetists (AAOP), Continuing Education Conference 3-91, Columbus, OH
Contact: AAOP National Office, (703) 836-7118*

September 6-8, 1991
2nd Scientific Meeting of the Scandinavian Medical Society of Paraplegia, Copenhagen, Denmark
Contact: Centre for Spinal Cord Injured, Rigshospitalet, TH2002, Blegdamsvej 9, DK-2100 Copenhagen, Denmark; Tel (+45)31 38 66 33, Ext. 2007

September 13-15, 1991
American Association for the Surgery of Trauma (AAST), Annual Meeting, Philadelphia, PA
Contact: AAST, c/o Dr. Lewis M. Flint Jr., SUNY, 462 Grider St., Buffalo, NY 14215

September 16-20, 1991
DUNDEE '91—International Conference and Instructional Course on Orthotics, Dundee, Scotland
Contact: Dundee '91 Secretariat, c/o Dundee Limb Fitting Centre, 133 Queen St., Broughty Ferry, Dundee, DD5 1AG, Scotland*

September 17-20, 1991
International Conference: Symbolic-Numeric Data Analysis and Learning, Paris, France
Contact: INRIA, Service des Relations Exterieures, Domaine de Voluceau - BP 105, 78153 Le Chesnay, Cedex, France; Tel (33.1) 39 63 55 01, Telex 697033F, Fax (33.1) 39 63 56 38

September 18-20, 1991
Dynamic Axial Fixation Course, Oswestry, Shropshire, UK
Contact: Erica Wilkinson, Symposium Secretary, The Robert Jones & Agnes Hunt Orthopaedic Hospital, Oswestry, Shropshire SY10 7AG, UK; Tel 0691 655311, Ext 3392*

September 22-25, 1991
International Meeting on Alzheimer's Disease, Amsterdam, The Netherlands
Contact: OBA, Europaplein, 1078 GZ, Amsterdam 020-5491212, The Netherlands

September 22-25, 1991
Meeting on Sensors and their Applications, Edinburgh, UK
Contact: The Meetings Office, Institute of Physics, 47 Belgrave Square, London SW1X 8QX, UK*

September 24-26, 1991
Eurospeech '91: 2nd European Conference on Speech Communication and Technology, Genova, Italy
Contact: Secretariat, Eurospeech '91, IIC-Instito Internazionale delle Comunicazioni, Via Pertnace-Villa Piaggio, 16125 Genova, Italy*

September 27, 1991
Transitional and Community Care of Patients with Neurological Disabilities, Symposium, Oswestry, Shropshire, UK
Contact: Erica Wilkinson, Symposium Secretary, The Robert Jones & Agnes Hunt Orthopaedic Hospital, Oswestry, Shropshire SY10 7AG, UK; Tel 0691 655311 Ext 3392*

September 27-28, 1991
Biomedical Engineering Society, Annual Fall Meeting, Charlottesville, VA
Contact: Biomedical Engineering Society, PO Box 2399, Culver City, CA 90231

September 28, 1991
Initial Total Care of the Patient with Spinal Injuries, Symposium, Oswestry, Shropshire, UK
Contact: Erica Wilkinson, Symposium Secretary, The Robert Jones & Agnes Hunt Orthopaedic Hospital, Oswestry, Shropshire SY10 7AG, UK; Tel 0691 655311 Ext 3392*

September 29-October 4, 1991
Medart International: First World Congress on Arts Medicine - Arts and Medicine, Medicine for Artists, Arts as Medicine, The Hague, The Netherlands
Contact: Hoboken Congress Organisation, Erasmus University, PO Box 1738, 3000 DR Rotterdam, The Netherlands; Tel 010-408 7881/7882*

October 1-6, 1991
American Orthotic and Prosthetic Association (AOPA), Annual National Assembly, Anaheim, CA
Contact: AOPA National Headquarters, 717 Pendleton St., Alexandria, VA 22314; (703) 836-7116

October 8-11, 1991
Sixth British Course on Knee Instability, Oswestry, Shropshire, UK
Contact: Erica Wilkinson, Symposium Secretary, The Robert Jones & Agnes Hunt Orthopaedic Hospital, Oswestry, Shropshire SY10 7AG, UK; Tel 0691 655311 Ext 3392*

October 13-16, 1991
7th International Conference on Mechanics in Medicine and Biology, Ljubljana, Yugoslavia
Contact: ICMMB 91, Technical Organiser, CANKARJEV DOM, Cultural and Congress Centre, Kidricev Park 1,61000 Ljubljana, Yugoslavia

October 13-16, 1991
15th Symposium: Computer Applications in Medical Care, Washington, DC
Contact: George Washington University Medical Center, Office of Continuing Education, 2300 K St. NW, Washington, DC 20037*

October 19, 1991
UK Intensive Care Society Annual Meeting, London, UK
Contact: The Honorary Secretary, Intensive Care Society, 9 Bedford Square, London WC1B 3RA, UK*

October 21-23, 1991
Combined Meeting of the Orthopaedic Research Societies of USA, Japan, and Canada, Banff, Alberta, Canada
Contact: Madeleine Aldridge, Conference Office, University of Calgary, 2500 University Dr. NW, Calgary, Alberta T2N 1N4, Canada; (403) 220-7319; Fax (403) 289-7287

October 22-24, 1991
International Robots and Vision Automation Show and Conference, Detroit, MI
Contact: Tel (313) 994-6088, Fax (313) 994-3338

October 23-27, 1991
Association of Rehabilitation Nurses (ARN), Annual Meeting, Kansas City, MO
Contact: ARN, 2506 Gross Point Rd., Evanston, IL 60201

October 27-November 1, 1991
American Academy of Physical Medicine and Rehabilitation (AAPM&R), Washington, DC
Contact: AAPM&R, 78 East Adams St., Suite 1300, Chicago, IL 60603

October 28-30, 1991
New Hampshire Chapter of American Physical Therapy Association, "Biomechanical Evaluation and Treatment of Foot and Ankle Dysfunction," Concord, NH
Contact: Diane Spahn, (603) 437-1026, Tuesdays through Fridays

October 28-30, 1991
Prescription Footwear Association (PFA), Annual Symposium, "Management of the Active Foot," Las Vegas, NV
Contact: PFA, 9861 Broken Land Pkwy., Suite 255, Columbia, MD 21046; Tel (800) 673-8447*

November 4-8, 1991
Acoustical Society of America (ASA), Houston, TX
Contact: ASA, 335 East 45th St., New York, NY 10017

November 4-14, 1991
Seminar on Computer-Aided Diagnosis, Warsaw, Poland
Contact: Professor Dr. Maciej Nalecz, Director, International Centre of Biocybernetics, 00-818 Warszawa, ul. Krajowej Rady Narodowej 55, Poland*

November 13-15, 1991
ACCESS EXPO, Conference and Exposition of Accessibility, Washington, DC
Contact: The Fairfield Factor, Inc., Access Expo Public Relations Counsel, 13 Obtuse Rocks Rd., Brookfield, CT 06804; Art Kerley, (203) 775-0422

November 15-18, 1991
National Rehabilitation Association (NRA), Annual Meeting, Louisville, KY
Contact: NRA, 633 South Washington St., Alexandria, VA 22314

November 17-20, 1991
15th Symposium on Computer Applications in Medical Care (SCAMC), Washington, DC
Contact: The George Washington University Medical Center, Office of Continuing Education, 2300 K St. NW, Washington, DC 20037

November 17-22, 1991
Joint Meeting of the American Association of Physicists in Medicine (AAPM) with the Radiological Society of North America, Chicago, IL
Contact: AAPM, 335 East 45th St., New York, NY 10017; (212) 661-9404

November 22-25, 1991
American Speech-Language-Hearing Association (ASHA), Annual Convention, Atlanta, GA
Contact: ASHA, 10801 Rockville Pike, Rockville, MD 20850; (301) 897-5700

November 22-25, 1991
Gerontological Society of America 44th Annual Science Meeting, San Francisco, CA
Contact: Jenny Youngdahl, Gerontological Society of America, 1275 K St. NW, Suite 350, Washington, DC 20005; Tel (202) 842-1275*

December 1-6, 1991
Joint Meeting of American Association of Physicists in Medicine (AAPM) with the Radiological Society of North America (RSNA), Chicago, IL
Contact: AAPM, 335 East 45th St., New York, NY 10017; (212) 661-9404

December 3-8, 1991
American Academy of Neurological and Orthopaedic Surgeons, Annual Meeting, Las Vegas, NV
Contact: American Academy of Neurological and Orthopaedic Surgeons, 2320 Rancho Dr., Suite 108, Las Vegas, NV 89102

December 5-6, 1991
Third International Symposium on Computer Simulation in Biomechanics, Perth, Western Australia
Contact: Ms. Rosemary Ingham, 3rd WGCS Symposium Secretariat, Dept. of Human Movement, The University of Western Australia, Nedlands, WA 6009, Australia*

December 6-10, 1991
AJUTEC'91—1st International Exhibition of Technical Aids and New Technologies for Disabled Persons, Matosinhos, Porto, Portugal
Contact: Divisao de Feiras e Exposicoes, Exponor—4450 Matosinhos, Porto, Portugal; Tel (02) 9957091/116/041/066, Telex 28751, Fax (02) 9957499*

December 9-13, 1991
13th International Congress on Biomechanics, Perth, Western Australia
Contact: 13th Congress Secretariat, Dept. of Human Movement, The University of Western Australia, Nedlands WA 6009, Australia

1992

February 20-25, 1992
American Academy of Orthopaedic Surgeons (AAOS) Annual Meeting, Washington, DC
Contact: AAOS, (312) 823-7186

March 6-7, 1992
American Orthotic and Prosthetic Association (AOPA), Region IV Meeting, Raleigh, NC
Contact: Don Ferguson, CPO, (919) 966-4630*

April 7-12, 1992
American Academy of Orthotists and Prosthetists (AAOP), Annual Meeting and Scientific Symposium, Miami, FL
Contact: AAOP National Office, (703) 836-7118

April 22-25, 1992
Independence 1992: International Congress and Exposition on Disability, "Independence: Self-Determination by Persons with Disabilities," Vancouver, BC, Canada
Contact: B.C. Pavillon Corp., 200-1190 Melville St., Vancouver, British Columbia VGE 3WI, Canada; (604) 689-5084, Fax (604) 689-4806*

May 14-16, 1992
American Orthotic and Prosthetic Association (AOPA), Region V Annual Meeting, Kings Island, OH
Contact: AOPA, 717 Pendleton St., Alexandria, VA 22314; (703) 836-7116

June 28-July 3, 1992
International Society for Prosthetics and Orthotics (ISPO), Seventh World Congress, Chicago, IL
Contact: ISPO Seventh World Congress, Moorevents, Inc., 400 N. Michigan Ave., Suite 2300, Chicago, IL 60611; (312) 644-5997*

July 12-15, 1992
Shoulder Surgery, Fifth International Conference, Paris, France
Contact: Convergences-V^{e}, ICSS, 120, avenue Gambetta, 75020 Paris, France; Fax +33 1 40 31 01 65*

July 19-24, 1992
International Arthroscopy Congress, Platja d'Aro, Spain
Contact: Ramon Cugat, MD, C.H.A.T., Paseo de Gracia, II. 08007, Barcelona, Spain; Tel 34 3 301 6500, Fax 34 3 302 0243

August 5-8, 1992
American Orthotic and Prosthetic Association (AOPA), Quad Regional Meeting, C'oeur D'Alene, ID
Contact: Jack Meredith, CO, (509) 326-6401*

August 9-14, 1992
4th International Congress of the Hard of Hearing, Jerusalem, Israel
Contact: Secretariat, PO Box 50006, Tel Aviv 61500, Israel; Tel 972 3 654571, Telex 341171, Fax 972 3 655674

August 23-27, 1992
American Association of Physicists in Medicine (AAPM), 34th Annual Meeting with the Division of Medical and Biological Physics of the Canadian Association of Physicists, Calgary, Alberta, Canada
Contact: AAPM, 335 East 45th St., New York, NY 10017; (212) 661-9404

September 7-11, 1992
17th World Congress of Rehabilitation International: Accelerating Efforts to Equalisation of Opportunities—Strategies for the 90s, Nairobi, Kenya
Contact: The Association for the Physically Disabled of Kenya Headquarters, Lagos Rd., PO Box 46747, Nairobi, Kenya; Tel 24443 and 332227*

September 14-18, 1992
XI World Congress of the International Federation of Physical Medicine and Rehabilitation, Dresden, GDR
Contact: Prof. Jurgen Kleditsch, Dept. of Physical Therapy and Research, Clinic of Orthopaedics, Medical Academy "Carl Gustav Carus," Dresden, GDR-8019

October 26-31, 1992
American Orthotic and Prosthetic Association (AOPA), Annual National Assembly, Orlando, FL
Contact: Annette Suriani, (703) 836-7116*

October 29-November 1, 1992
14th Annual International Conference of the IEEE EMBS, Paris, France
Contact: Swamy Laxminarayan, Academic Computing Center, MSB-A539, NJ Medical School, 185 South Orange Ave., Newark, NJ 07103, or Jean Louis Coatrieux, Lab Traitement du Signal, Universite de Rennes I, Campus de Beaulieu, Rennes Cedex, France*

November 8-11, 1992
16th Symposium on Computer Applications in Medical Care (SCAMC), Baltimore Convention Center, Baltimore, MD
Contact: The George Washington University Medical Center, Office of Continuing Education, 2300 K St. NW, Washington, DC 20037

November 20-23, 1992
American Speech-Language-Hearing Association (ASHA) Annual Convention, San Antonio, TX
Contact: Frances Johnston, ASHA, 10801 Rockville Pike, Rockville, MD 20852; (301) 897-5700

November 29-December 4, 1992
Joint Meeting of the American Association of Physicists in Medicine (AAPM) with the Radiological Society of North America (RSNA), Chicago, IL
Contact: AAPM, 335 East 45th St., New York, NY 10017; (212) 661-9404

1993

February 18-23, 1993
American Academy of Orthopaedic Surgeons (AAOS) Annual Meeting, San Francisco, CA
Contact: AAOS, (312) 823-7186

March 30-April 4, 1993
American Academy of Orthotists and Prosthetists (AAOP), Annual Meeting and Scientific Symposium, Las Vegas, NV
Contact: AAOP, (703) 836-7116

June 10-12, 1993
7th Congress of the European Society for Shoulder and Elbow Surgery, Aarhus, Denmark
Contact: Orthopaedic Hospital, Randersvej 1, DK-8200 Aarhus N. Denmark; Tel +45 86 16 75 00 Ext 4622, Fax +45 86 10 77 33

August 8-12, 1993
35th Annual Meeting of the American Association of Physicists in Medicine (AAPM), Washington, DC
Contact: AAPM, 335 East 45th St., New York, NY 10017; (212) 661-9404

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October 12-16, 1993
American Orthotic and Prosthetic Association (AOPA) Annual National Assembly, Reno, NV
Contact: Annette Suriani, (703) 836-7116*

November 19-22, 1993
American Speech-Language-Hearing Association (ASHA) Annual Convention, Anaheim, CA
Contact: Frances Johnston, ASHA, 10801 Rockville Pike, Rockville, MD 20852; (301) 897-5700*

November 28-December 3, 1993
Joint Meeting of the American Association of Physicists in Medicine (AAPM) with the Radiological Society of North America (RSNA), Chicago, IL
Contact: AAPM, 335 East 45th St., New York, NY 10017; (212) 661-9404

1994

April 9-16, 1994
IRMA VII - The Seventh World Congress of the International Rehabilitation Medicine Association: 25th Anniversary of IRMA, Washington, DC
Contact: IRMA VII, 875 Kings Hwy., West Deptford, NJ 08096*

July 24-28, 1994
American Association of Physicists in Medicine (AAPM), 36th Annual Meeting, Anaheim, CA
Contact: AAPM, 335 East 45th St., New York, NY 10017; (212) 661-9404

November 18-21, 1994
American Speech-Language-Hearing Association (ASHA) Annual Convention, Washington, DC
Contact: Frances Johnston, ASHA, 10801 Rockville Pike, Rockville, MD 20852; (301) 897-5700*

November 27-December 2, 1994
Joint Meeting of the American Association of Physicists in Medicine (AAPM) with the Radiological Society of North America (RSNA), Chicago, IL
Contact: AAPM, 335 East 45th St., New York, NY 10017; (212) 661-9404

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VA REHABILITATION DATABASE

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